Gait Characteristics Change Following an Acute Exposure to Kneeling and Filtering Considerations for Gait Analysis

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Abstract

Following sustained kneeling in young adults, kinetic and kinematic changes at the knee and ankle during gait have been observed (Gaudreault et al., 2013; Kajaks & Costigan., 2015a; Tennant et al., 2014). It is possible that a change in the cyclic gait pattern after sustained deep knee bending (greater than 120 degrees of flexion) could differentially load the cartilage in the knee and modify the mechanical stress exerted on the tissues, at least acutely, leading to increased risk of knee osteoarthritis development (Edd et al., 2018; Kajaks & Costigan, 2015). Gait changes that have been observed in those with knee osteoarthritis include decreases in knee flexion at heel strike, knee flexion range of motion, knee power, ankle power and knee flexion moment, and increases in knee adduction moment, and vertical loading rate. The primary purpose of this thesis was to explore whether differences in gait kinematics and kinetics following sustained kneeling mirror those occurring in osteoarthritic gait. It was hypothesized that prolonged kneeling would increase external peak knee adduction moment (KAM), and vertical loading rate (VLR), and would decrease knee flexion angle at heel strike (HS), knee flexion range of motion (ROM), external peak knee flexion moment (KFM) in early stance, positive peak ankle power (PAP) in late stance, and the second positive peak knee power (PKP) in late stance.

There is also a need to low-pass filter marker data to remove random noise, especially because noise is amplified when differentiating position and orientation for the calculation of linear and angular velocities and accelerations for inverse dynamics calculations such as joint moments and joint powers (Kristianslund et al., 2012; Sinclair et al., 2013). It has become common practice to filter raw marker and raw ground reaction force (GRF) data differently to preserve ground reaction force peaks and impact due to heel strike while walking. Papers reviewed for this project include a range of marker cut off frequencies from 5-8Hz and a range of GRF cut-off frequencies from raw (no filtering) to 40Hz. Selecting different cut-off frequencies for marker and GRF data can result in large oscillations in joint moments

around heel strike during running that are not representative of the actual movement (Bezodis et al., 2013). It is possible that these effects could exist in any form of gait. The secondary purpose of this thesis was to determine the effects of various low-pass filter cut-offs on all dependent variables calculated for the primary objective. It was hypothesized that large differences in the filter cut-offs between marker and GRF data would result in significantly different knee flexion angles at HS, knee flexion ROM, first peak KAM, first peak KFM, PAP, PKP, and peak VLR.

As part of a previous study, fourteen participants (8M/6F) performed three pre- and post-kneeling gait trials at their self-selected pace while motion (Optotrak, NDI, Waterloo, Canada) and ground reaction force (AMTI, Watertown, MA, USA) data were recorded for the right leg. The kneeling protocol consisted of three ten-minute bouts of sustained plantarflexed kneeling separated by five-minutes of seated rest. Ground reaction force and motion capture data were filtered at 6Hz using a 4^{th} order Butterworth filter using Visual 3D software (C-Motion Inc., Germantown, MD) to determine outcome measures. For the primary objective, seven two-tailed paired t-tests (one per outcome measure) were performed to compare subject mean values pre- and post- kneeling ($\alpha=0.05$). For the secondary objective, marker and GRF data were filtered using a 4th order zero-lag Butterworth filter with 16 different cuff-off combinations. Each combination of GRF and marker input data was then used to calculate KFM, KAM, PAP and PKP for each subject's pre- and post- kneeling gait trials. For measures where only kinematic data or only ground reaction forces were involved in their calculation (knee angle at HS knee ROM and peak VLR), only six different cut-off frequencies were used. A two-way repeated measures ANOVA was used to compare the mean outcome variables across different filtering conditions, and between the pre- and post-kneeling time points. Any main effect of filtering condition would indicate that filtering condition had a significant effect on the outcome measure. Any interaction would indicate that the effect of the kneeling exposure (part of the primary objective) depended on the filtering condition.

From the primary objective analysis, acute exposure to kneeling produced significant increases in knee flexion angle at HS, and peak KFM. An increase in knee flexion angle at HS and KFM peak (which typically occurs during early stance) suggests that participants could be attempting to reduce knee joint rate of loading more by accepting their weight with more knee flexion, as opposed to more knee extension which is typical of osteoarthritic gait and is related to greater axial loading rate at the knee. The increase in external KFM is balanced by the internal knee extensor muscle moment, suggesting that following kneeling, contradictory to what we hypothesized, participants are placing greater demand on their quadriceps. High flexion activities such as kneeling are associated with an increased risk of knee OA development. This work suggests that prolonged kneeling has the potential to compromise the integrity of the knee joint such that kneeling can acutely alter the loading patterns experienced at the knee joint (and thus potentially other lower limb joints) during the subsequent ambulation.

For the secondary objective of this thesis there was a significant main effect between pre- and post-kneeling for knee flexion at HS ($p=0.034$), and KFM ($p=0.0063$). The significant main effects between pre- and post-kneeling conditions for knee flexion at HS and KFM were expected, these results were found when investigating the primary objective. Interestingly there was also a main effect for filtering condition of peak VLR ($p=0$). The raw filtering condition was significantly larger than all other conditions, followed by the 25Hz being larger then 6Hz and 10Hz, and finally 20Hz being significantly larger then 6Hz. Since there was no significant main effect for filtering condition for any of the remaining dependent variables used, we can conclude that these peak measures are robust to filtering condition. With the lack of filter x kneeling condition interactions for all variables, we can infer that pre- and post-differences between subjects discreate variables are also robust to filtering condition. The lack of significant filtering condition effect for all outcome measures but the peak VLR is likely due to these measures being far enough away from heel strike, that the additional noise from oscillations caused by the impact at heel strike, like that seen during running, had minimal effect on these peaks except for peak VLR. This work shows that, in young asymptomatic participants, post-kneeling changes in kinetic and kinematic outcomes commonly used in the osteoarthritis literature are robust to changing filter cut-off frequencies. This finding may suggest that the wide variety of cut-off frequencies (when reported at all) in the literature on gait characteristics in osteoarthritis may not be of significant concern when comparing these outcome measures between studies.

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1 Introduction

Previous studies have indicated that high knee flexion activities (>120°), such as kneeling and squatting, are associated with increased risk of knee osteoarthritis (OA) development (Coggon et al., 2000; Herquelot et al., 2015; Palmer, 2012; Pejhan et al., 2020; Schram et al., 2020; Verbeek et al., 2017). The adaptations reported in a population of occupational kneelers and in individuals exposed to an acute kneeling exposure largely resemble quadriceps avoidance gait seen in OA populations (Gaudreault et al., 2013; Hunt et al., 2020; Kajaks & Costigan, 2015a; Kline et al., 2019; Tennant et al., 2014). It is possible that a change in the cyclic gait pattern after sustained deep knee bending could differentially load the cartilage in the knee and modify the mechanical stress exerted on tissues, leading to an increased risk of knee OA development (Edd et al., 2018a; Favre & Jolles, 2016; Gaudreault et al., 2013; Hunt et al., 2020; Kajaks & Costigan, 2015). Changes, such as abnormal loading, in gait due to kneeling and squatting postures can tamper with the homeostatic state of the knee joint and lead to cartilage damage and knee OA (Edd et al., 2018b; Kajaks & Costigan, 2015; Pejhan et al., 2020). Workers that reported kneeling or squatting at work were 1.7 times more likely to develop knee OA than workers who reported not kneeling or squatting at work (Verbeek et al., 2017). Previous findings have indicated that there is an increased risk of developing symptomatic knee OA in a group of floor layers with a substantial amount of kneeling work (Jensen et al., 2012).

In osteoarthritic gait, knee moments, joint power, knee range of motion, ground reaction forces, and muscle activity have been shown to differ from healthy norms. The external knee adduction moment (KAM) during the stance phase of gait is a robust measure, is often used to identify osteoarthritic gait (Kumar et al., 2015), and has often been used as a surrogate for medial compartment contact forces (Zhao et al., 2007). More recent work suggests KAM is more closely related to the ratio of medial to lateral loading in the joint (Kutzner et al., 2013; Meyer et al., 2013). KAM peak and impulse have also been shown to be strong predictors of knee OA initiation and progression (Calder et al., 2014; Chehab et al., 2014; D'Souza et al., 2022; Davis et al., 2019; Erhart-Hledik et al., 2021). In the sagittal plane, the external knee flexion moment (KFM) is reduced during midstance in individuals with knee OA as an attempt to reduce articular loads. This measure is often related to the reduction in flexion angle excursion seen during gait (Favre & Jolles, 2016; Gaudreault et al., 2013). Knee extensor power reflects the speed with which the joint moment can be developed to move or support the knee (McGibbon & Krebs, 2002; Segal et al., 2009). Individuals with knee OA present with a "stiff gait" where their knees tend to be more extended through out stance. This includes less knee flexion during heel strike which has been observed to coincide with an increase in peak vertical loading rate, and less range of motion throughout gait (Al-Zahrani & Bakheit, 2002; Farrokhi et al., 2015; Mündermann et al., 2005).

The way in which these outcome measures are processed varies from study to study. Marker data is typically low-pass filtered to remove random noise, especially because noise is amplified when differentiating orientation for the calculation of angular velocities and acceleration for inverse dynamics calculations such as joint moments and subsequent calculations of joint powers (Kristianslund et al., 2012; Sinclair et al., 2013). It has become common to see filtering cut-offs in the OA gait literature ranging between 5-8Hz for marker data and raw (no filtering cut-off) to 40Hz for GRF data (Table 2.1). The selection of different cut-off frequencies may be an attempt to preserve as much of the impact of heel strike as possible. A difference in cut-off frequencies for marker and GRF data can result in large oscillations in joint moments around heel strike during running that are not representative of the actual movement and its possible that these effects could exist in any form of gait (Bezodis et al., 2013).

The goal of this research is to examine the effects of an acute kneeling exposure on gait parameters of young healthy individuals. An increase in knee adduction moment (KAM) and larger amounts of variation in knee flexion and adduction moments have previously been reported following a

30 minute kneeling exposure in young adults (Kajaks & Costigan, 2015; Tennant et al., 2018). Given that individuals with continuous kneeling exposure are at greater risk of developing knee OA and some characteristics of OA gait are present following acute kneeling exposures it is reasonable to theorize that the adoption of OA characteristics alter joint loading thus eventually beginning progression of the disease. The secondary goal of this study was to determine the effects of various low-pass filter cut-offs on commonly reported measures in the knee OA literature. This work will allow us to contribute to existing models of high knee flexion and gait, in addition to contribute to the literature on the mechanical pathway of knee OA. Specific objectives and hypotheses are detailed in Table 4.1.

2 Literature Review:

2.1 Incidence and severity of knee OA

Osteoarthritis ranks among the top 10 causes of disability world-wide, and is associated with significant pain and stiffness, fatigue, and activity limitations (Gignac et al., 2020). It is characterized by progressive cartilage loss, remodeling of adjacent bones and concomitant local low-grade inflammation (Georgiev et al., 2019). Knee OA is a mobility limiting, age related disease for which there is a 44% lifetime risk in American adults and OA represents a huge socioeconomic burden worldwide (Edd et al., 2018; Murphy et al., 2008). There is no known cure for OA, making prevention efforts for maintaining an individual's functional capacity into old age a priority (Segal et al., 2009).

2.2 Risk factors associated with knee OA

There are multiple risk factors that can contribute to the development of knee OA including repetitive kneeling, previous injury at the knee, older age, being female heavy lifting and being overweight/obese or having a history of obesity (BMI>30) (Anderson et al., 1988; Herquelot et al., 2015; Linnan et al., 2017; Manninen et al., 2002; Palmer, 2012; Schram et al., 2020). Previous studies have indicated that high knee flexion activities (>120°), such as kneeling and squatting, are associated with increased risk of knee OA development (Coggon et al., 2000; Herquelot et al., 2015; Palmer, 2012; Schram et al., 2020; Verbeek et al., 2017). A cumulative kneeling exposure of 1-2 hours or 30 cycles of high flexion postures/workday has been associated with increased risk of knee OA in occupational settings (Awosoga et al., 2019; Coggon et al., 2000; Laudanski et al., 2019; Linnan et al., 2017). Odds ratios have been used to express the likelihood of knee OA development for individuals who have been exposed to high flexion in the workplace for >1 hour or >30 cycles per day compared to matched controls who were not exposed to occupational kneeling or squatting. For individuals who kneel for >1hour/day in total, males had an OR of 1.7 and females 2.0. For individuals who squat for >1 hour/day in total, males had an OR of 2.2 and females 2.8. For those who get up from kneeling or squatting >30 times/day, males had an OR of 2.0 and females 1.8. (Coggon et al., 2000). When comparing knee OA risk in kneeling and the duration of exposure, individuals who reported kneeling for >1 hour/day for 1-9.9 years and >20years have OR of 3.0 for men 2.8 for women and 1.7 in men and 3.9 in women respectively (Coggon et al., 2000). For getting up from kneeling or squatting >30 times/day 1-9.9 years compared to >20 years produced OR of 2.1 for men 2.0 for women and 2.3 for men and 3.9 for women respectively (Coggon et al., 2000). The OR from kneeling and squatting were the largest compared to any other workday activity measured (Coggon et al., 2000). When considering occupational stress factors, the high mechanical stress at the knee joint due to kneeling and squatting indicate these activities are modifiable risk factors for the development of knee OA (Pejhan et al., 2020; Verbeek et al., 2017).

2.3 Possible underlying mechanisms

Both mechanisms explained below are based on the idea that knee OA is caused by frequent abnormal loading, resulting in cumulative stress beyond the tissues tolerance limits. The modifiable risk factors of frequent high flexion activities daily such as kneeling, have been repeatedly published. However, there is still little information concerning the biomechanical mechanisms responsible for initiation of knee OA and how they are related to these high-flexion activities.

2.3.1 Integrated joint system model for OA progression

Gait adaptations reported in a population of occupational kneelers and in populations exposed to an acute kneeling exposure largely resemble OA gait, which has also been described as “quadriceps avoidant” or “stiff” gait (Gaudreault et al., 2013; Hunt et al., 2020; Kajaks & Costigan, 2015a; Kline et al., 2019; Tennant et al., 2014). It is possible that a change in the cyclic gait pattern after sustained deep knee bending could differentially load the cartilage in the knee and modify the mechanical stress exerted on tissues, leading to OA-like gait characteristics and increased risk of knee OA development

(Edd et al., 2018; Favre & Jolles, 2016; Gaudreault et al., 2013; Hunt et al., 2020; Kajaks & Costigan, 2015b). Kneeling has also been associated with 65% greater odds of Hoffa synovitis (inflammation of the infrapatellar fat pad) compared to individuals that did not kneel for 30 minutes during a single day. If they did this for even just one day a week, they had significantly greater odds of prevalent Hoffa synovitis on an MRI (Van Ginckel et al., 2019). Although Hoffa synovitis readings are known for being non-specific, it's likely that kneeling compressed the Hoffa fat pad and induced the inflammation to a point where individuals may begin to compensate (Van Ginckel et al., 2019). One recent mechanism of disease progression has been described extensively by Edd et al. (2018) whose proposed integrated joint system model includes gait mechanics, subchondral bone mineral density, and cartilage thickness in a homeostatic state in healthy individuals (Figure 2.1).

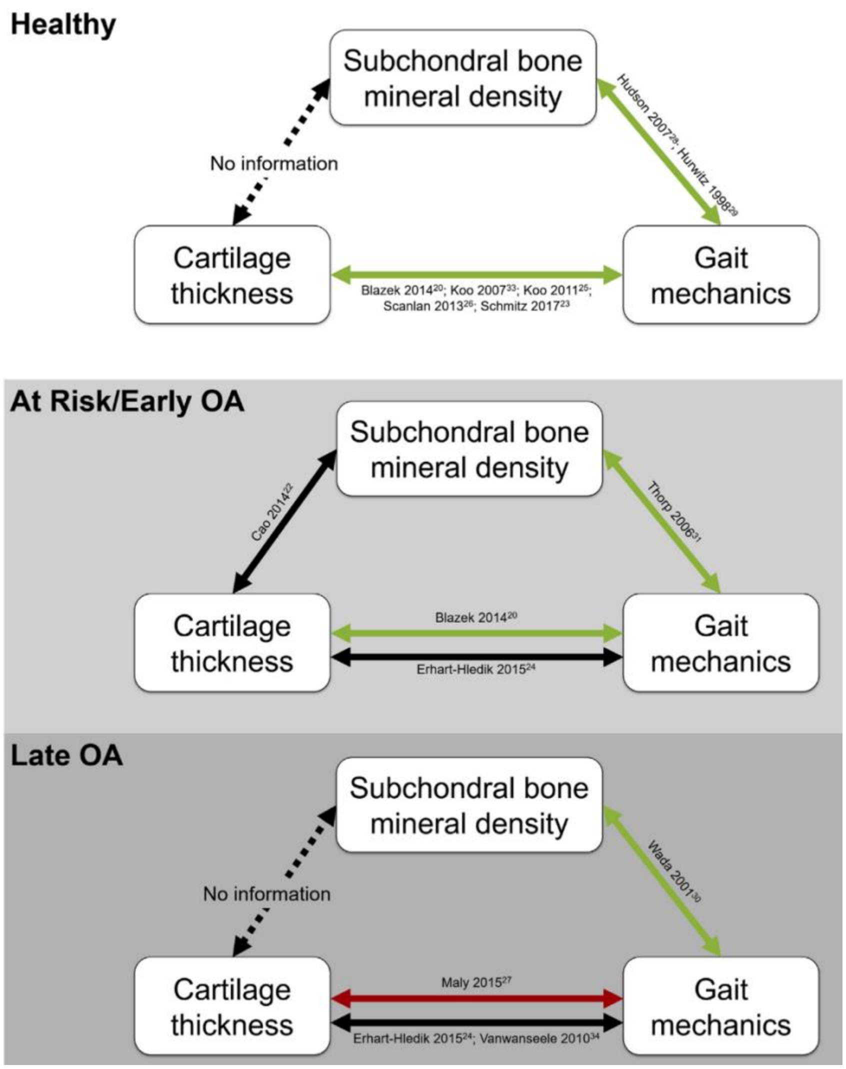

Figure 2.1 Summary of correlations between cartilage, subchondral bone mineral density, and gait mechanics for three stages of knee health. Solid line arrows indicate the correlations that have been tested in literature, solid line arrows indicate the strength and sign of the reported correlations: green indicates moderate to strong (|r|>0.3) positive correlations, red indicates moderate to strong (|r|>0.3) negative correlations, and black indicates weak (|r|<0.3) or inconclusive correlations. The dashed line arrows correspond to correlations that have not yet been tested.[1]

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[1] Reprinted from Osteoarthritis and Cartilage, 26/11, Edd, S. N., Omoumi, P., Andriacchi, T. P., Jolles, B. M., & Favre, J. Modeling knee osteoarthritis pathophysiology using an integrated joint system (IJS): a systematic review of relationships among cartilage thickness, gait mechanics, and subchondral bone mineral density,1425-1427. Copyright (2018) with permission from Elsevier

In healthy knees all the properties of the integrated joint system (gait, subchondral bone mineral density and cartilage thickness) are adapted to each other where, for example, an increase in biomechanical load during gait increases cartilage thickness. A knee remains healthy so long as the properties are in this homeostatic state where they are adaptive to each other (Edd et al., 2018). Subchondral bone mineral density is positively corelated to the KAM and KAM impulse (largest KAM coincided with the largest subchondral bone mineral density on the medial tibial plateau), so it is likely that these two measures can be used to represent the relative loading between the medial and lateral compartments (Edd et al., 2018). This relationship stays consistent with all stages of knee health (Edd et al., 2018).

Once homeostasis has been disrupted, possibly through any of the previously mentioned risk factors like mechanical stress on the tibiofemoral joint from an individual's occupation, the relationship between properties changes, reflecting relationships present in individuals with OA. It appears that at least some of these relationships change as the disease progresses. In groups classified as at-risk/early OA knees the relationships between knee joint properties and gait measures are less apparent then those reported in healthy and late OA knees (Blazek et al., 2014; Edd et al., 2018; Erhart-Hledik et al., 2021; Thorp et al., 2006). It is speculated that it is at this time the relationships are shifting and may be harder to detect. In late OA knees, there is a negative correlation between KAM and KAM impulse and cartilage thickness (Edd et al., 2018; Erhart-Hledik et al., 2021; Maly et al., 2015; Vanwanseele et al., 2010; Wada et al., 2001). Because there are minimal reported moderate or strong relationships between the integrated joint system in risk/early OA knees, detection of the initiation of OA has been elusive and the objective of much research in the OA community. Changes such as abnormal loading in gait due to kneeling and squatting postures could modify the homeostatic state and lead to cartilage damage and knee OA (Edd et al., 2018; Kajaks & Costigan, 2015b; Pejhan et al., 2020; Tennant et al., 2018).

2.3.2 *Model of potential mechanism of high knee flexion*

A secondary model describing specifically how kneeling could be associated with abnormal loading of the non-contractile structures at the knee and has been previously published by Kajaks and colleagues (2015). This model has two potential mechanisms (Figure 2.2). One, where deep-knee flexion alters the location of tibia-femoral contact location and increases contact pressure on the cartilage and two, where the posterior displacement and external rotation of the tibia relative to the femur during deep knee flexion places additional stress on the knee joint ligaments (PCL and MCL) (Kajaks & Costigan, 2015). It is hypothesized that both mechanisms alter joint control, which can be detected during ambulation following kneeling (Kajaks & Costigan, 2015).

Higher knee flexion angles are associated with higher external flexion moments at the knee joint during double and single leg ascent and decent during kneeling and squatting (Nagura et al., 2001; Pejhan et al., 2020; Thambyah, 2008). Plantarflexed kneeling like that used in the Kajacks study elicits the largest knee flexion moment during the descent, ascent and static phases (Pejhan et al., 2020). In addition to the large knee joint contact forces and contact stress that already occur during high knee flexion movement (Kingston & Acker, 2020; Nagura et al., 2006), it has been reported that, following prolonged static kneeling, there is an increase in variation of the knee flexion moment and knee adduction moment with no consistent direction of variation (Kajaks & Costigan, 2015). It's possible that static kneeling could acutely alter the integrity of the viscoelastic structures within the knee joint and that this is what is responsible for the observed altered gait pattern(Kajaks & Costigan, 2015).

It is also possible that the mechanisms in which static kneeling and dynamic kneeling affect the joint are different. Previous work has also demonstrated that, following a more dynamic kneeling exposure where individuals repeated cycles of 2 minutes of single arm supported kneeling followed by 30s of full flexed kneeling for 30 minutes, individuals exhibited significantly greater peak KAM, and delayed onset of vastus medialis in subsequent ambulation. Following 30 minutes of rest peak KAM returned to baseline and the delayed onset of vastus medialis remained (Tennant et al., 2018). Regardless of how high flexion alters gait, there is evidence to show that gait changes are associated with kneeling and it's known that abnormal gait mechanics are a potential mechanism for knee OA development (Andriacchi et al., 2004; Felson, 2004).

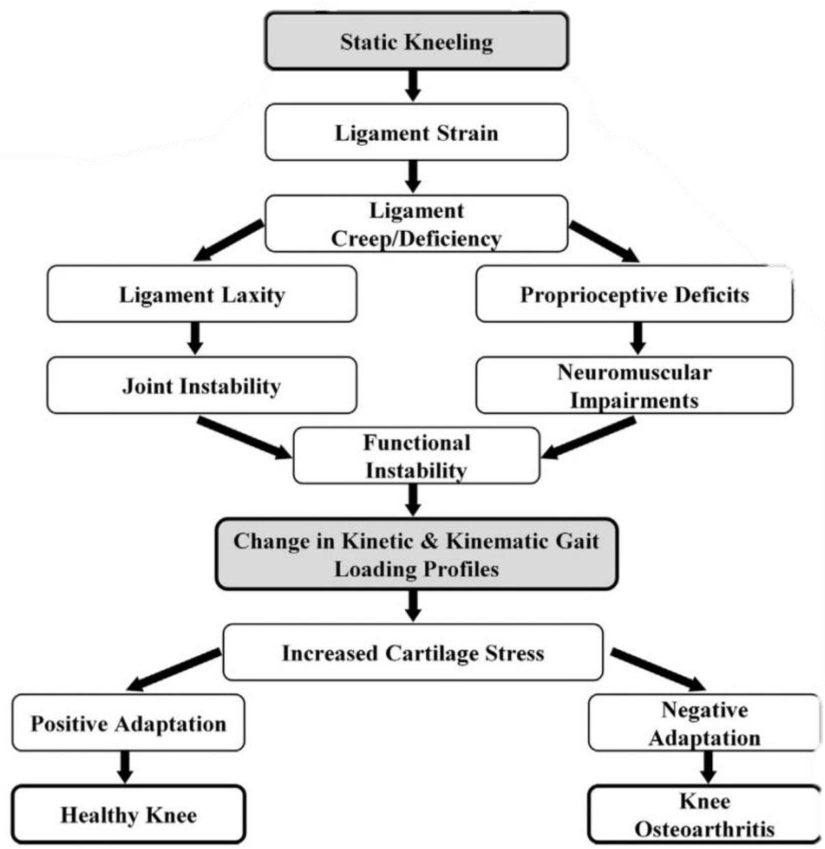

Figure 2.2 A hypothesized model of the relationship between static kneeling and knee[2]

[2] Reprinted from Applied Ergonomics, 46/Part A, Kajacks, T., & Costigan, P. The effect of sustained static kneeling on kinetic and kinematic knee joint gait parameters,224-230.,Copyright (2015) with permission from Elsevier

2.4 Biomechanical variables associated with Knee OA

2.4.1 Knee joint moments

The external knee adduction moment (KAM) has often been used as a surrogate for medial tibiofemoral compartment contact forces (Zhao et al., 2007), however somewhat more recent work (Walter et al., 2015) has contradicted the idea that the KAM is a measure of medial compression. Instead, KAM is more closely related to the ratio of medial to lateral loading in the joint (Kutzner et al., 2013) . Regardless, KAM has been shown to be a strong predictor of knee OA initiation and progression (Chehab et al., 2014; D'Souza et al., 2022; Davis et al., 2019; Erhart-Hledik et al., 2021). A recent meta-analysis reported that individuals with higher early stance peak KAM during a 24-12 month follow had 1.88 greater odds of OA progression in the medial tibiofemoral joint using a Kellgren Lawrence grade, bone marrow lesions and cartilage damage/deficits (D'Souza et al., 2022). One study suggests that an increase of 5% BW*Ht is sensitive to predicting OA progression over 6 years, this is one of the only studies that provided a threshold value to evaluate progression risk (Miyazaki et al., 2002). Conversely, another study found that changes in knee adduction moment during gait were associated with all individuals with knee OA regardless of severity suggesting that it may have limited ability to discriminate between severity levels (Deluzio & Astephen, 2007). It should be noted that Deluzio & Astephen (2007) used clinical criteria that defined moderate OA as individuals diagnosed with radiographic OA but not eligible for knee replacement surgery and defined severe OA as subjects scheduled for knee replacement surgery immediately following the gait testing, whereas Miyazaki et al (2002) used Kellgren and Lawrence grade and joint space narrowing to identify disease progression. As knee OA is a progressive disease and difficult to classify, it's likely that groups overlap in biomechanical patterns, meaning that walking patterns are not entirely defined by severity level.

Studies that have examined the relationship between KAM, knee flexion moment (KFM) and medial compartment loading (Kutzner et al., 2013; Walter et al., 2015) also suggest that the addition of peak KFM strengthens the relationship with medial compartment loading. Where gait strategies that decrease KAM but increase KFM often present no change in medial compartment compression and being able to reduce both KAM and KFM is correlated with a decrease in medial compartment compression (Walter et al., 2015). Surrogate measures for knee joint loading are used due to the complexity of accurately estimating internal joint loading during movement, with non-invasive methods. While greater knee extensor moment during gait may lead to enhanced lower extremity stability via greater quadriceps activity, the associated higher joint compressions may also result in greater pain and put the articular cartilage at higher risk for further erosion (Farrokhi et al., 2015).

2.4.2 Knee joint angles

Previously the gait kinematics of workers exposed to knee straining postures (kneeling and squatting) and non-knee straining postures were compared during treadmill walking (Gaudreault et al., 2013). Individuals exposed to knee straining postures had an increased knee flexion angle at heel strike, and a decreased flexion angle range of motion, when compared to controls (Gaudreault et al., 2013). Another study on lower limb loading found that all patients with knee OA landed (first foot contact) with the knee in a more extended position and experienced a more rapid increase in the GRF (higher loading rate)(Mündermann et al., 2005). Self-reported knee instability was associated with decreased knee motion (sagittal plane) during weight acceptance and midstance (Farrokhi et al., 2015). Knee flexion excursion during the early stance phase of gait provides “shock absorption” through eccentric quadriceps muscle activity, reduction in knee motion excursion can have negative consequences such as increased impulse loading, axial compression and potentially accelerated rates of disease progression (Farrokhi et al., 2015).

Individuals with knee OA have less mean flexion, during loading and swing phases (Al-Zahrani & Bakheit, 2002). A previous systematic review exploring knee mechanics of healthy and knee OA knees using medical imaging outcomes demonstrated that osteoarthritic knees have a more adducted orientation throughout flexion and have a more anterior contact pattern in the lateral compartment (Scarvell et al., 2018). It is possible that the change in flexion excursion throughout gait is related to the change in contact patterns. It is unclear on weather this is adaptive in an attempt to offload areas of cartilage or maladaptive and overloading areas of cartilage (Scarvell et al., 2018).

2.4.3 Knee Power

Knee extensor power reflects the speed with which the joint moment can be developed to move or support the knee (McGibbon & Krebs, 2002; Segal et al., 2009). In general, peak power represents energy from the muscles that is either being added to the body (positive power) to sustain the upright posture and forward progression or taken out of the body (negative power) to slow segment motion or cushion directional changes due to walking (Winter, 2009)**.** Power differences across the lower limb joints reflect differences in energy demands that are placed on muscles crossing the joints (Winter, 2009).

In individuals with knee OA, the ability to produce a larger knee flexion moment at toe off and greater positive power at toe-off was associated with higher function as measured by a 400-m walk test (Segal et al., 2009).. Individuals that compensate with an ankle strategy where they disproportionately increase positive ankle power and decrease positive knee power at toe-off have been identified in healthy and higher functioning patients with knee OA (Segal et al., 2009). In men, sagittal hip energy absorption during terminal stance and ankle energy generation during pre-swing were measures that distinguished between high and low-functioning participants (McGibbon & Krebs, 2002; Segal et al., 2009). A study done by Calder et al. (2014) revealed that increased knee extensor power showed a positive relationship with medial compartment loading during level walking for individuals with clinical knee OA, where those capable of producing large knee powers also demonstrated large KAM peaks and impulses (Calder et al., 2014). It is possible that the ability of the knee to control and transfer energy from the ankle to the hip and pelvis was tied to increased functional abilities of individuals.

Subtle changes in gait function may not be identifiable without examining the mechanistic source of such changes. Analysing the effects of interrelated movements of several body segments and joints can be difficult and confusing. Using a simplified measure such as joint power which combines angular velocity and moments has potential to identify compensations more easily (Winter, 2009). Diminished ankle power at push-off is one characteristic that appears to be a generalized trait of disablement and perhaps of aging alone (McGibbon & Krebs, 2004). It has previously been reported that differences in concentric ankle mechanical energy expenditure between groups could not be explained completely by differences in gait speed (McGibbon & Krebs, 2004). Therefore, low functioning patients with knee OA reduce ankle push off power for reasons other than diminished walking speed (McGibbon & Krebs, 2002, 2004).

In healthy individuals, prior to push-off the knee normally undergoes a second positive peak in power generation. This is where energy is transferred distally and may contribute to the energy that is ultimately delivered to the foot at toe-off (McGibbon & Krebs, 2002). In individuals with OA there is a lack of the second peak due to early knee extension, reducing knee angular acceleration and minimal energy transfer due to a reduced knee flexion moment (McGibbon & Krebs, 2002). Individuals with OA also present with a higher concentric compensation co-efficient at the knee in paced gait, meaning all energy entering one segment is delivered from the adjoining segment (ankle), requiring no muscular assistance (McGibbon & Krebs, 2002). This suggests that patients with knee OA reduce knee loading and agonist muscle contractions to limit the loads experienced by the joint. This finding is consistent with the association between quadriceps weakness and knee OA. However, it is unclear on whether these patients had weak quadriceps or just avoided using them (McGibbon & Krebs, 2002). Since power has never been calculated following a kneeling exposure or in populations of chronic kneelers, it is worth measuring to detect subtle coping mechanisms in gait following kneeling. Although there is not enough data to definitively hypothesize a direction of change, it seems reasonable to think that, following kneeling, individuals may adapt in a similar manner to individuals with knee OA, where they may attempt to avoid loading the knee by increasing the demand on the ankle.

2.4.4 Ground Reaction Forces and Vertical Loading Rate

Unlike some of the aforementioned variables like joint moments, impulse and power, ground reaction forces can be collected with relative ease, require less computational processing and display similar patterns during gait as joint moments (Costello et al., 2021; Mündermann et al., 2005; Winter, 2009). However, reported GRF outcome measures vary in the OA literature. The components of the GRF (lateral and vertical) have been related to OA severity, while some gait changes seem to be related to increased vertical loading rate (slope of the GRF curve) immediately following heel strike during walking (Mündermann et al., 2005). Although, there is a lack of consensus in the literature on the best protocol for determining ground reaction force variables, which is likely due to the uncontrolled variation in walking speeds during assessments or variations in how researchers control for walking speed (Astephen Wilson, 2012). Since ground reaction force is affected by gait speed, and individuals with OA often present with slower gait speeds as a protective mechanism, it makes it difficult to separate the two variables (Costello et al., 2021; Wiik et al., 2017). However recently a study on cross sectional differences in dynamic GRF patterns during walking found that some differences in GRF patterns remained after adjusting for speed (Costello et al., 2021). For example differences in medial-lateral GRF patterns remained regardless of speed adjustment and less dynamic vertical GRF in individuals with radiographic OA and pain also remained (Costello et al., 2021).

It has previously been reported that all patients with OA hit the ground in a more extended position (moderate OA vs controls 1.8 deg vs 5.3 deg) (Mündermann et al., 2005). Striking the ground in more extension coincides with a reported increase in vertical loading rate of ground reaction force (Mündermann et al., 2005). Since the cartilage in the knee is viscoelastic the rate at which it is loaded matters. This increase in vertical ground reaction force loading rate has been reported to coincide with an increase in axial loading rate at the ankle, knee, and hip (Mündermann et al., 2005). This increased loading rate may potentially cause more rapid cartilage degeneration at the hip, knee, and ankle (Costello et al., 2021; Mündermann et al., 2005). In addition, a more rapid increase in ground reaction force indicates a more rapid shift of the body's weight from the contralateral limb to the support limb (Mündermann et al., 2005). This rapid shift is proposed to be a potential mechanism of gait compensation to reduce the mediolateral distance between the centre of mass and the knee joint centre (Mündermann et al., 2005). Thus reducing the moment arm of the GRF and potentially reducing the peak KAM (Mündermann et al., 2005).

2.5 Filtering Considerations for Gait Analysis

The way in which gait outcomes are processed varies from study to study. Marker data is typically low-pass filtered to remove random noise, especially because noise is amplified by successive differentiation of displacement data to calculate various outcomes (Kristianslund et al., 2012; Pezzack et al., 1977; Sinclair et al., 2013). Its seems likely that these effects could exist in any form of gait where there is an impact on a force plate (Bezodis et al., 2013). Although there are some methods for choosing cut-off frequencies such as a harmonics analysis, a residual analysis of the difference between filtered and unfiltered signals over a wide range of cut off frequencies (Winter, 2009; Yu et al., 1999), or equations (e.g. Yu et al., 1999) to estimate the optimal cut-off frequency, the rationale for the choice of cut-off frequency (and often the cut-off frequencies themselves) are not typically reported. Where only one of the 32 papers listed below reported performing a residual analysis (Schmitz & Noehren, 2014). Gait papers relating to knee OA that have been reviewed for this literature review, reported ranges of raw marker cut-off of 5-8Hz and raw ground reaction force cut-offs from no filtering at all to 40 Hz (Table 2.1).

Table 2.1 Of the 32 papers on OA gait that were read for this project, the 13 that reported a filtering cut-off for marker data or GRF data are presented bellow alongside with the gait outcome measures. N/A means that the cut-off was not reported.

Author	Outcomes	Marker filter cut-off	GRF filter cut-off
Zhao et al., 2007	KAM Total knee joint contact force Medial knee joint contact force	6Hz	N/A
Maly et al., 2013	KAM KAM impulse	6Hz	N/A
Robbins et al., 2011	KAM KAM impulse	6Hz	N/A
Freisinger et al., 2017	Frontal plane excursion	6Hz	N/A
Madadi-Shad et al., 2019	Knee and Ankle angles GRF peaks, Time to peak GRF, GRF impulse GRF loading rate	6Hz	20Hz
Schmitz & Noehren, 2014	KAM Knee adduction angle Centre of pressure, Vertical and lateral GRF	8Hz	35Hz
Calder et al., 2014	KAM KAM impulse	6Hz	N/A
Resende et al., 2012	Sagittal hip, knee, and ankle power	6Hz – filtered power wave forms	N/A
Segal et al., 2009	Hip, knee, and ankle power Hip, knee, and ankle moments	6Hz	10Hz
Tennant et al., 2018	KAM Vertical rate of loading	6Hz	Raw -Moments 100Hz - vertical rate of loading
Kajaks & Costigan, 2015	Knee flexion and adduction angle KFM and KAM	6Hz	25Hz
Farrokhi et al., 2015	Hip, knee, and ankle sagittal angles Hip, knee, and ankle moments Vertical GRF	6Hz	40Hz
Costello et al., 2021	Vertical, lateral, and anterior-posterior GRF	N/A	Raw

3 *Rationale*

OA develops after a long period of repetitive exposure. Multiple studies have indicated that frequent high flexion postures like squatting and kneeling in occupational settings increase an individual's odds of developing knee OA. Despite the growing epidemiological evidence that frequent high knee flexion in an occupational setting place workers at a greater risk of knee OA, biomechanical support for the link between high flexion postures and knee OA is lacking. It's possible that the ambulation following kneeling involves alterations to and individual's cyclic gait pattern. This atypical gait pattern could alter the normal load on the cartilage and non-contractile tissue in the knee and modify the mechanical stress exerted on tissues, leading to osteoarthritic gait and increased risk of knee OA development (Edd et al., 2018; Kajaks & Costigan, 2015). For this reason, it is worthwhile to measure gait variables that change with knee OA presence and progression (such as knee joint angles, knee moments, knee power, ankle power and vertical loading rate) to determine what changes those outcomes undergo following kneeling. This rationale leads to the question: Following kneeling, do healthy adults adopt gait compensations like individuals with knee osteoarthritis?

The ways in which biomechanical signals are processed vary study to study. In some cases, filter cut-off frequencies are not reported, and many do not explain their rationale for selecting a certain filter cut-off frequency. It is important to determine the robustness and repeatability of commonly used gait measures so that they can be accurately compared to the existing literature and contribute to the ability to make larger conclusions by combining the results of multiple studies. This rationale leads to the question: How much impact does signal filtering have on commonly used reported measures of knee load?

4 *Objectives and Hypotheses*

The primary purpose of this study is to examine the effects of an acute kneeling exposure on gait parameters of young healthy individuals. An increase in knee adduction moment and larger amounts of variation in knee flexion and adduction moments have previously been reported following a 30 minute kneeling exposure in young adults (Kajaks & Costigan, 2015; Tennant et al., 2018). Given that individuals with continuous kneeling exposure are at greater risk of developing knee OA and some characteristics of OA gait are present following acute kneeling exposures, it is hypothesised that following kneeling there will be:

1. an increase in knee flexion angle at heel strike and greater knee flexion range of motion during stance
2. an increase in peak knee adduction moment and decrease in peak knee flexion moment during the first 50% of stance
3. a reduction in power generation at the knee (second positive peak knee power) and greater power generation at the ankle (peak positive ankle power) during the last 50% of stance
4. a change in peak vertical loading rate in the first 10% of stance

The secondary purpose of this study was to determine the effects of various low-pass filter cut-off frequencies on all the discrete variables calculated in the analysis to address the primary objective. It was hypothesised that the cut-off frequencies used to process marker and GRF data would have a significant effect on knee flexion at heel strike, knee flexion range of motion, first peak knee adduction moment, first peak knee flexion moment, peak positive ankle power, second peak knee power and peak vertical loading rate.

The primary and secondary research questions, corresponding hypotheses, and the rationale for each of the hypotheses are summarized in Table 4.1

Table 4.1 Questions, Hypothesis and Rationale

Questions	Hypothesis	Rationale
	The kneeling exposure will result in the following gait changes:	
Following kneeling, do healthy adults adopt gait compensations like individuals with knee osteoarthritis?	1. Increase in knee flexion angle at heel strike and greater knee flexion range of motion	Previous studies have shown that individuals that kneel in their occupation tend to have higher flexion angles and range of motions during gait compared to controls. However, individuals with knee OA tend to walk with more knee extension and less range of motion. These compensations are assumed to change the rate of loading and change the location of contact on the tibiofemoral cartilage during gait. It is possible that, following a single exposure to kneeling, there is an increase in knee flexion angle at HS and flexion ROM like that adopted in occupational kneelers.
	2. An increase in KAM and decrease in KFM during the weight acceptance phase of gait	Individuals with knee OA have been shown to walk with increased medial compartment loading represented by the peak knee adduction moment and the knee adduction impulse. During early stance (weight acceptance) some individuals with early stages of knee OA with increased peak knee adduction moment have compensated by decreasing the knee flexion moment. It is possible that we will see similar compensations following kneeling.
	3. A reduction in power generation at the knee and greater power generation at the ankle during the latter half of stance	It has previously been shown that the contributions of the hip, knee, and ankle in the propulsion phase of stance is different in individuals with knee OA. Females tend to adopt an ankle strategy by increasing their ankle power to minimize the contribution of the knee extensors. If this occurs in OA it is possible that this compensation exists in individuals following kneeling to offload structures around the knee.
	4. A change vertical loading rate	Individuals with medial knee OA walk with an increased vertical ground reaction force loading rate, assumed to be due to the more extended position of the knee during heel strike. Because GRF are key components to inverse dynamics it is worth looking at how the loading rates change following a kneeling exposure.

How much impact does signal filtering have on commonly used reported measures of knee load?	1. Cut-off frequencies used to process marker and GRF data would have a significant effect on knee flexion at heel strike, knee flexion range of motion, (KAM), first peak knee flexion moment (KFM), peak positive ankle power (PPAP), second peak knee power (PPKP), peak vertical loading rate(VLR).	While the effect of different cut-off frequencies has not been reported in walking gait, significant effects of cut-off frequencies on knee moment during running have been reported (Bezodis et al., 2013). It is important to recognize how robust and repeatable our commonly used measures are such as knee moments and power, so that they can be accurately compared to the existing literature and contribute to the ability to make larger conclusions based of repeatable results.

Table 4.2 A table containing the variables to be calculated, the associated abbreviations and a brief definition

Variable	Abbreviation	Definition
Knee flexion at heel strike	Knee flexion at HS	Knee flexion angle taken at heel strike (degrees)
Knee flexion range of motion	Knee flexion ROM	The knee flexion angle range during stance phase (degrees)
Knee adduction moment	KAM	Peak knee adduction moment during the first 50% of stance phase (Nm)
Knee flexion moment	KFM	Peak knee flexion moment during the first 50% of stance phase (Nm)
Peak positive ankle power	PAP	The peak positive ankle power that occurs in late stance (50-100%) (W)
Second peak positive knee power	PKP	The second peak positive knee power which occurs in late stance (50-100%)((W)
Vertical loading rate	VLR	The peak rate of loading from the first 10% of stance phase (0-50%) (N/s)

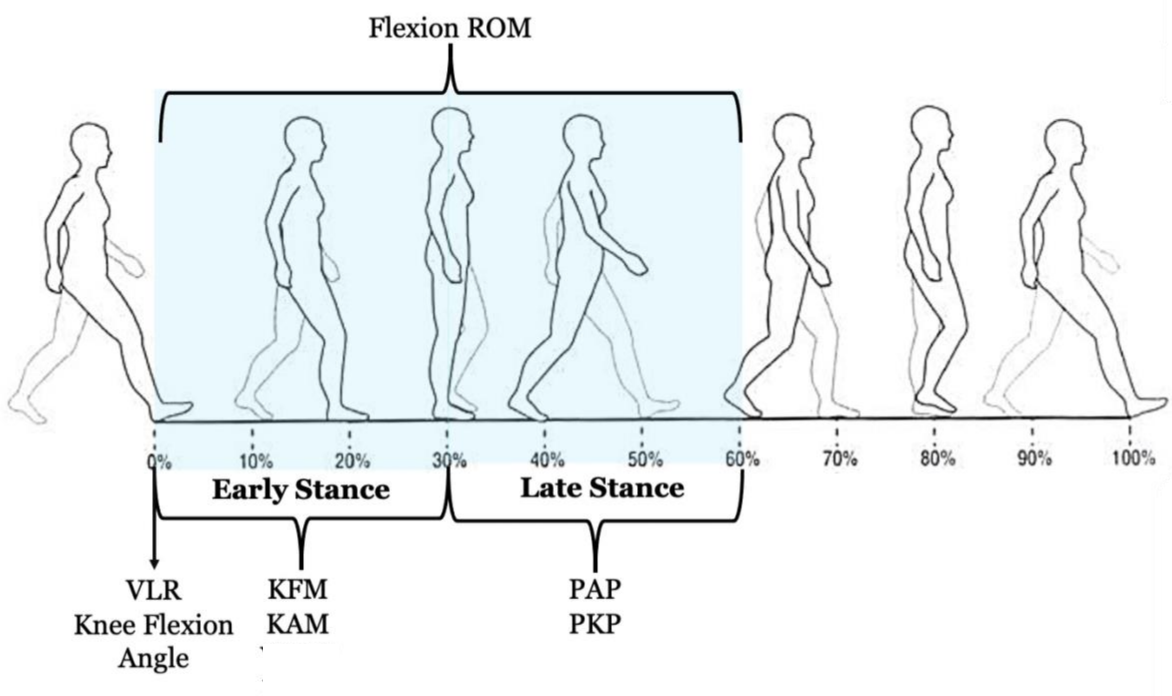

Figure 4.1 A visual representation of the timing of each of the variables during the gait cycle

5 Methods

Fourteen young healthy subjects were recruited (8M/6F, 22.4 years (+/- 2.5), (169.6 cm (+/- 8.5), 70.8 kg (+/- 12.8) for a previously completed thesis project focusing on the effects of kneeling on frontal plane knee joint laxity (Mines, 2016). Participants were excluded if they reported having current pain in the lower limbs, or previous lower limb injuries that required surgery, taking analgesics or anti-depressive medication. The age of participants was limited to 30 years as age is a potential factor in frontal plane laxity scores which was the initial purpose of this data (Mines, 2016). This study was reviewed and received clearance through the University of Waterloo Research Ethics Committee and all participants provided written informed consent.

Marker clusters, each equipped with 5 Optotrak smart markers in a non-collinear orientation, were placed on the participant bilaterally on the thighs and unilaterally on the right shank, and foot using straps, double sided tape and Hypafix medical tape to prevent marker slipping. The following landmarks were digitized to define segments: Thigh – greater trochanter, lateral epicondyle, and medial epicondyles of femur; Shank – lateral and medial epicondyles of femur, medial and lateral malleoli; Foot – lateral and medial malleoli, calcaneus, 1st, and 5th metatarsal heads. All digitization was performed with the participant standing in the anatomical position. Kinematic data was collected using 6 Optotrak position sensors (Northern Digital Inc., Waterloo, ON, Canada) at a sampling rate of 64 Hz. Kinetic data was collected with four AMTI force platforms (Advances Mechanical Technology Inc., Watertown, MA, USA) at a sampling rate of 2048 Hz. First principles and an Optotrak Data Acquisition Unit (ODAU II, Northern Digital inc., Waterloo, ON, Canada) with a multiplexer was used to synchronize marker and force data.

5.1 Gait Analysis and Kneeling Exposure:

Participants ambulated barefoot until three successful gait trials were collected, where the participant's right foot cleanly hit a force plate. Since this is a repeated measures design, the effects of gait velocity on outcome measures are less of a concern. Therefore, participants walked at their natural pace to capture the most natural gait possible. Following the initial gait trials, participants were asked to kneel with their buttocks as close to their heels as possible to induce the largest amount of knee flexion, and with the ankles plantarflexed. Foam padding was provided for the participants to place under their ankles as they assumed the kneeling posture to cushion the top of the foot and prevent discomfort. Thin mats were placed on top of the force plates to minimize the pain from knee-ground contact forces. Participants kneeled for 3 cycles of 10 minutes with 5 minutes of rest in between (Kajaks & Costigan, 2015). During rest periods, participants were seated on a chair and were asked to sit with little movement of the lower limbs. After the kneeling exposure, participants again completed gait trials until three successful trials were collected.

5.2 Data Processing:

Ground reaction force and motion capture data were processed using Visual 3D software (C-Motion Inc., Germantown, MD) and MATLAB (MathWorks, Inc., Natick, MA). Kinematic and ground reaction force data were padded with a reflection of 10 points on either end and filtered using a dual pass 4^{th} order Butterworth low-pass filter at a cut-off frequency of 6Hz (Howarth & Callaghan, 2009; Winter, 2009). Following filtering the padding was removed. The filtering equations used by Visual 3D can be found in (Appendix B). Missing data points were interpolated using a third-order cubic spline to fit up to 10 consecutive missing frames (Howarth & Callaghan, 2010). The link segment model created in Visual 3D used the segment coordinate systems in Table 5.1. A Euler rotation matrix of X, Y, Z (flexion/extension, abduction/adduction, internal/external rotation) was used to describe the relative three-dimensional orientation between segments. Power was calculated as a scalar term that is the dot product of the three-dimensional moment and angular velocity vectors.

Ankle power was calculated during stance from the ankle joint angular velocity, which is the derivative of the ankle joint angles with respect to time, and the ankle moment expressed in the foot coordinate system (Table 5.1). Knee moments were expressed about the axes of the tibial coordinate system. Knee power was calculated from the knee joint angular velocity, and knee joint moments resolved to the tibial coordinate system (Table 5.1). Body and segment mass and inertial parameters used in the inverse dynamics calculations were the default set in Visual 3D which are derived by using published anthropometric data with geometric modeling (Dempster, 1955; Hanavan, 1964; C-Motion Inc., Germantown, MD). All variables were time normalized to 100% of stance phase, defined at the first point >10N and the last point >10N which is standard practice in Visual 3D (C-Motion Inc., Germantown, MD). Positive external moments in the sagittal plane represent flexion moments, and in the frontal plane represent adduction moments, expressed in the tibial co-ordinate system. For the calculation of knee adduction moment impulse, non-normalized knee adduction moments were integrated using the trapezoidal method (Calder et al., 2014; Maly, 2008). The time derivative of vertical ground reaction force was calculated in each frame for the first 10% of the gait cycle and the peak of these derivatives was used as the peak vertical loading rate (Mündermann et al., 2005).

Table 5.1 Segment co-ordinate systems used for the inverse dynamics calculations of the lower limb (Pejhan et al., 2020). Planes are fit to the points listed in the plane descriptions.

Segment	Description
Foot	Origin: Midpoint between lateral and medial malleoli XZ plane: Lateral and medial malleoli; 1st and 5th metatarsal X axis: Vector through the midpoints of lateral and medial malleoli and 1st and 5th metatarsals pointing anteriorly Y axis: Perpendicular to the XZ plane Z axis: Cross product of X and Y
Tibia	Origin: Midpoint between lateral and medial edges of the tibial plateau, projected onto the lateral-medial functional knee joint axis YZ plane: Lateral and medial edges of the tibial plateau, lateral and medial malleoli Y axis: Vector through the midpoints of the edges of the tibial plateaus and malleoli pointing superiorly X axis: Perpendicular to the YZ plane, projecting anteriorly Z axis: Cross product of X and Y
Thigh	Origin: Hip joint centre (Bell et al., 1989) YZ plane: Greater trochanter, lateral and medial femoral epicondyles Y axis: A vector between the hip joint centre and the midpoint of the femoral medial and lateral epicondyles, pointing superiorly X axis: Perpendicular to the YZ plane, projecting anteriorly Z axis: Cross product of X and Y

To address the secondary objective of this thesis project, analysis on this data included filtering the marker and GRF data using a 4th order zero-lag Butterworth filter at five different cut-off frequencies: 6Hz, 10Hz, 15Hz, 20Hz, and 25Hz in Visual 3D software (C-Motion Inc., Germantown, MD).The combinations used for comparison can be found in Table 5.2 in addition to non-filtered raw data. Each combination of GRF and marker input data was then used to calculate peak KFM, peak KAM, PAP and PKP for each subject's pre- and post- kneeling gait trials. Only the filtering conditions in the grey boxes (Table 5.2) in addition to raw data were used for the calculation knee flexion angle at HS, knee flexion ROM, and peak VLR. (There are no combinations of different cut-off frequencies because these outcomes require marker data only or force data only.)

Table 5.2: The cut off frequency combinations used for the secondary analysis. The grey boxes are conditions where the GRF and motion data were filtered at the same frequency. Combinations are referred to in the format of MiFj where i and j represent the cut-off frequencies applied to the marker(M) and GRF (F) data.

		Cut off frequency for ground reaction force data				
		6Hz	**10Hz**	**15Hz**	**20Hz**	**25Hz**
Cut off frequency for motion data	**6Hz**	6Hz	M6F10	M6F15	M6F20	M6F25
	10Hz		10Hz	M10F15	M10F20	M10F25
	15 Hz			15Hz	M15F20	M15F25
	20 Hz				20Hz	M20F25
	25Hz					25Hz

5.3 Statistical Analysis:

From each trial, knee flexion at HS, knee flexion ROM during stance, peak KFM from early stance, peak KAM from early stance, PKP from late stance, PAP from late stance, and peak VLR from the first 10% of stance were extracted (Table 4.2). Events were distinguished based on the percentage of stance time in which they occurred after normalizing to 100% of stance time. Subject dependent variables were averaged across pre-kneeling and across post-kneeling trials. All participants' mean distributions fulfilled the assumption of normality. To address the first objective, two tailed paired samples t-tests on participant means were used to test for differences between pre-and post-kneeling outcome variables. Alpha was set to 0.05 prior to conducting the experiment. For the secondary objective, a two-way repeated measures ANOVA was used to compare the mean outcome variables across different filtering conditions, and between the pre- and post- time points. Since the t-test used to address objective 1 tested for differences between pre- and post- kneeling measures, the main outcome from the ANOVA that addressed objective 2 was whether a significant main effect of filtering condition was detected. The potential for an interaction was explored to determine if any significant changes in gait outcomes determined in the t-tests for objective 1 would be affected by the filtering conditions. Since only the filtering main effect and interaction effects were of interest, post-hoc analyses were only carried out on those effects, if found to be significant.

6 Results

6.1 Gait Analysis and Kneeling Exposure

The t-tests indicated that, following the kneeling exposure there was a significant increase in knee flexion angle at heel strike (t=2.35, df=13, p=0.0354,Figure **6.1**), and peak knee flexion moment (t=-2.73, df=13, p=0.0171, Figure 6.2). No significant differences were found for flexion range of motion (t=0.1073, df=13, p=0.9162) (Figure **6.1**), peak knee adduction moment (t=-1.82, df=13, p=0.0918) (Figure 6.2), second peak positive knee power (t=-0.56, df=13, p=0.5837) (Figure 6.3), peak ankle power (t=-0.99, df=13, p=0.3386) (Figure 6.3), and peak vertical loading rate (t=-0.40, df=13, p=0.6974) (Figure **6.4**) between pre and post kneeling (Table **6.1**). Figures 6.1-6.7 show the mean change between pre and post kneeling gait trials for each subject for each dependent variable listed in Table **6.1**.Appendix Figure C.1 to C.3 display the mean wave forms for all subjects with the standard deviation shaded for the following dependent variables: knee flexion angle (Appendix Figure D.1), KFM, KAM (Appendix Figure D.2), PAP, and PKP (Appendix Figure D.3). Following the kneeling protocol knee flexion angle at HS and peak KFM by 1.13° (+/- 1.79°), 3.87Nm (+/- 5.30Nm). Post-hoc, the minimal detectable difference (MDD) and standard error of measurement (SEM) was calculated to ease with the interpretation or the results (Harvill, 1991).

Table 6.1: Paired t-test results in addition to the pre- and post-kneeling grand means, the mean difference within subjects (MD) with standard deviations in brackets and the minimal detectible difference (MDD) with the standard error of measurement (SEM) that was calculated on the non-normalized pre-kneeling gait data.

Variable	Pre-Mean (SD)	Post-Mean (SD)	MD (SD)	P -value	MDD (SEM)
Knee Flexion at Heel Strike (degrees)	-5.7 (5.3)	-6.8 (5.1)	1.6 (1.8)	0.0354*	6.0 (2.2)
Flexion ROM (degrees)	38.1 (3.7)	38.0 (2.4)	0.1 (3.0)	0.9162	8.3 (3.0)
Knee Flexion Moment Peak (Nm)	32.3 (16.1)	36.2 (16.2)	3.9 (5.3)	0.0171*	15.4 (5.5)
Knee Adduction Moment Peak (Nm)	25.7 (9.0)	27.1 (9.3)	1.4 (2.9)	0.0918	7.3 (2.6)
Second Peak Positive Knee Power (W)	30.1 (15.2)	30.23 (17.5)	1.1 (11.5)	0.9732	17.5 (6.3)
Peak Ankle Power (W)	173.9 (57.4)	183.6 (48.9)	5.8 (48.6)	0.4292	73.9 (26.7)
Peak Vertical Loading Rate (N/s)	45.6 (12.7)	42.3 (17.2)	0.6 (11.8)	0.3246	11.7 (4.2)

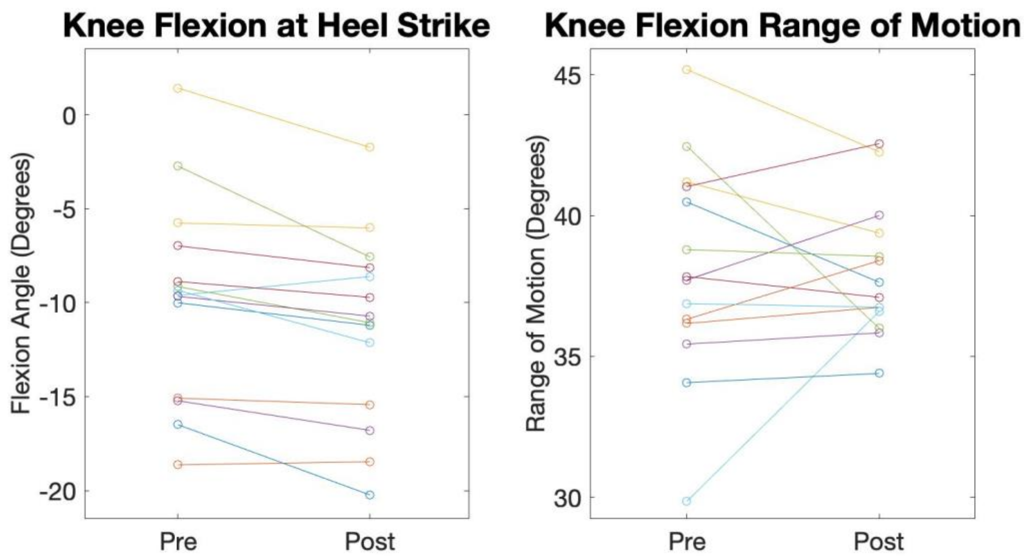

Figure 6.1 Each subject's mean knee flexion angle at heel strike and mean knee flexion range of motion during stance phase for the pre- and post-kneeling gait trials. Following kneeing there was a significant increase in knee flexion angle at heel strike.

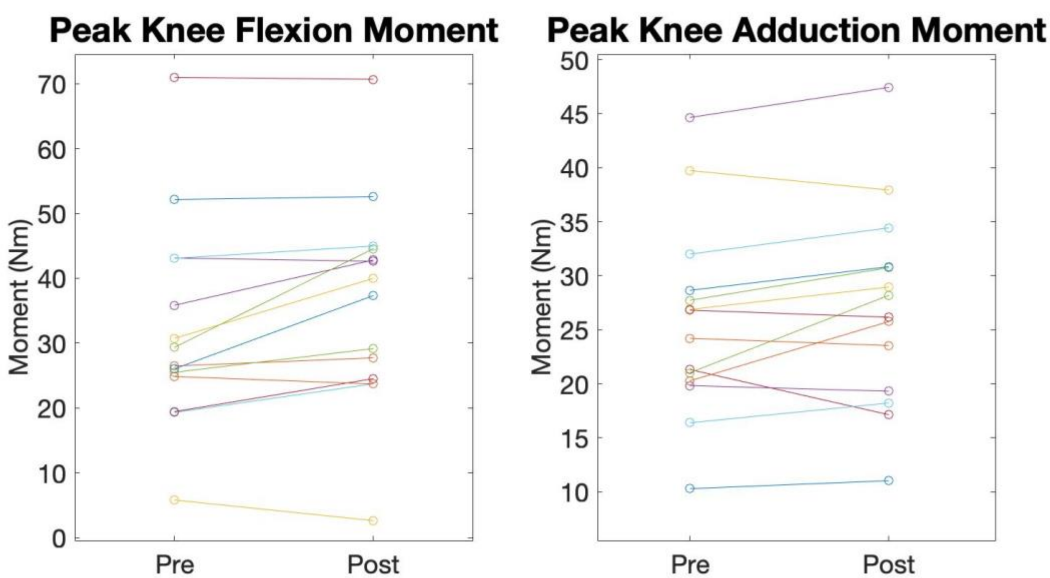

Figure 6.2 Each subject's mean peak knee flexion moment and mean peak adduction moment for the pre- and post-kneeling gait trials. Following kneeling there was a significant increase in peak knee flexion moment.

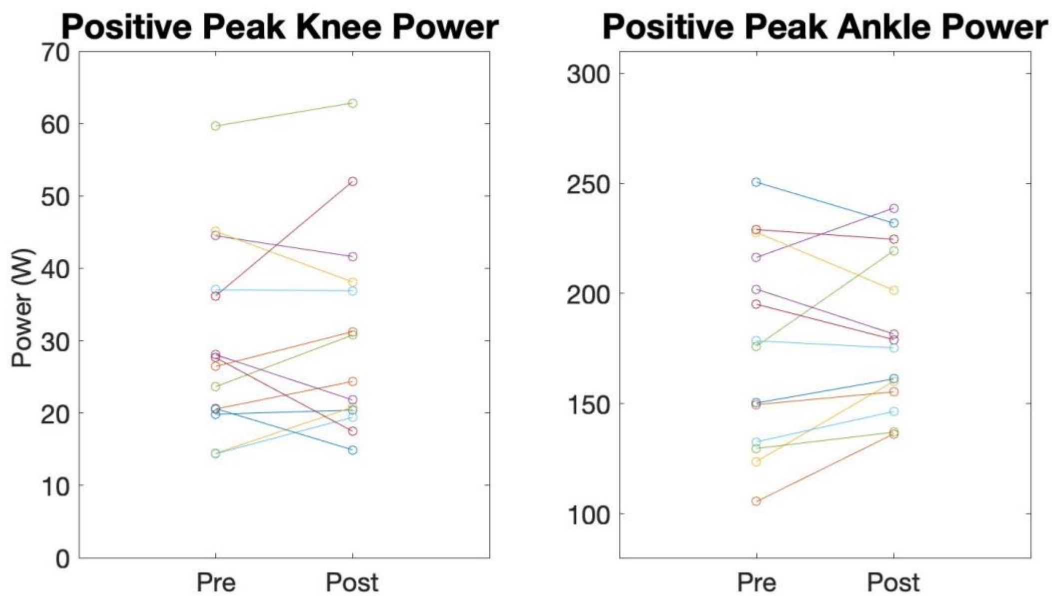

Figure 6.3 Each subjects mean second positive peak knee power from the late half of stance phase and mean peak ankle power for the pre- and post-kneeling gait trials.

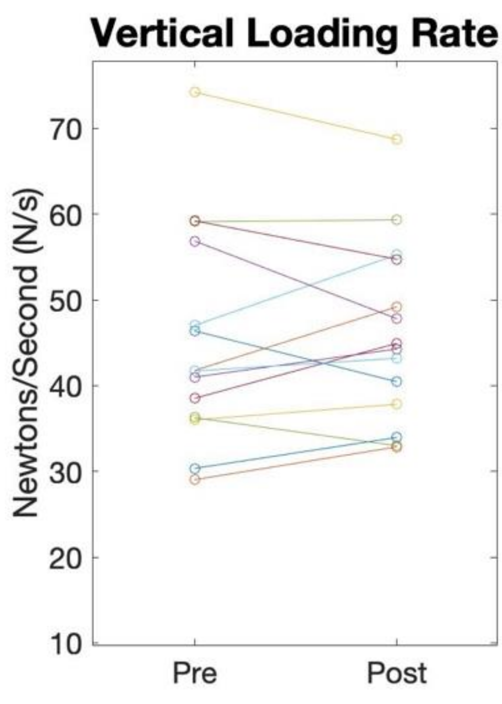

Figure 6.4 Each subject's mean max vertical loading rate for pre- and post-kneeling gait trials.

6.2 Gait Analysis and Filtering

Ten two-way repeated measures ANOVAs were conducted to determine the effect of filtering condition, on subjects pre- and post-kneeling gait measures including knee flexion angle at HS, knee flexion ROM, peak KFM, peak KAM, PKP, PAP, and peak VLR. Although the main purpose of the ANOVA was to evaluate the main effect of filtering to address Objective 2 and the filtering condition x kneeling condition interaction (as an additional analysis to determine if differences between kneeling conditions would be affected by filtering conditions), there was a significant main effect between pre- and post-kneeling for knee flexion at HS ($F4.53_{(1,156)}$, $p=0.035$) **Appendix Table C.**1 and KFM ($F=5.56_{(1,416)}$, $p=0.019$) **Appendix Table C.**4.

There was no significant main effect for filtering condition for knee flexion at HS ($F=0.1_{(5,156)}$, $p>0.05$), knee flexion ROM ($F=0.09_{(5,156)}$, $p>0.05$), KAM ($F=0.02_{(15,416)}$, $p>0.05$), KFM ($F=0.03_{(15,416)}$, $p>0.05$), PAP ($F=0.1_{(15,416)}$, $p=1$), PPKP ($F=0.14_{(15,416)}$, $p>0.05$). There was a significant main effect for filtering condition for peak VLR ($F=21.44_{(5,156)}$, $p=0$) where the 20Hz, 25Hz and Raw filtering condition were significantly greater than the 6Hz condition. The 25Hz and Raw filtering condition were significantly greater than the 10Hz condition and the Raw condition was significantly greater than all other filtering conditions (**Table** 6.5)

There were no significant filtering condition x kneeling condition interactions. Filtering condition did not significantly change the peak values for knee flexion angle at HS, knee flexion ROM, KAM, KFM, PAP, PKP. Filtering condition does play a significant role in the magnitudes of peak VLR. However, filtering condition also did not significantly change the difference between pre- and post-kneeling differences in any of the discrete variables we measured (Appendix **C**).

Table 6.2 pre and post kneeling means and standard deviations across participants for all filtering conditions used in the two way-way repeated measures ANOVAs for peak knee adduction moment (KAM), KAM impulse, and peak knee flexion moment (KFM). Filtering combinations are referred to in the format of MiFj where i and j represent the cut-off frequencies applied to the marker(M) and GRF (F) data. The last row ("Range") displays the minimum and maximum values across all filtering conditions.

Filter	Peak KAM (Nm)		Peak KFM (Nm)	
	Pre	Post	Pre	Post
6Hz	25.7 (7.0)	27.1 (9.3)	32.3 (16.1)	36.2 (16.2)
10Hz	26.2 (9.3)	27.6 (9.4)	32.8 (16.2)	36.9 (16.2)
15Hz	26.3 (9.2)	27.8 (9.4)	32.7 (15.9)	36.9 (16.1)
20Hz	26.4 (9.1)	27.9 (9.4)	32.8 (16.0)	37.0 (16.1)
25Hz	26.5 (9.0)	28.1 (9.2)	33.6 (16.1)	38.5 (17.5)
M6F10	25.9 (9.1)	27.4 (9.3)	32.4 (16.5)	36.3 (16.4)
M6F15	26.1 (9.1)	27.5 (9.3)	32.3 (16.5)	36.2 (16.4)
M6F20	26.2 (9.2)	27.5 (9.3)	32.3 (16.5)	36.1 (16.3)
M6F25	26.2 (9.2)	27.6 (9.3)	32.3 (16.5)	36.1 (16.3)
M10F15	26.3 (9.3)	27.6 (9.4)	32.8 (16.3)	36.8 (16.1)
M10F20	26.3 (9.3)	27.6 (9.4)	32.7 (16.3)	36.7 (16.1)
M10F25	26.3 (9.3)	27.6 (9.3)	32.7 (16.3)	36.7 (16.1)
M15F20	26.4 (9.2)	27.8 (9.4)	32.8 (15.8)	36.9 (16.0)
M15F25	26.4 (9.2)	27.9 (9.4)	32.8 (15.8)	37.0 (15.9)
M20F25	26.4 (9.1)	27.9 (9.4)	32.8 (16.0)	36.9 (16.2)
Raw	26.4 (9.4)	27.6 (9.4)	32.9 (15.8)	36.5 (15.6)
Range	25.7-26.4	27.1-28.1	32.3-32.9	36.2-37.0

Table 6.3 Pre and post kneeling means and ranges across participants for the 16 different filtering combinations used in the two-way repeated measures ANOVA for second positive peak knee power and peak positive ankle power. Filtering combinations are referred to in the format of MiFj where i and j represent the cut-off frequencies applied to the marker(M) and GRF (F) data. The last row ("Range") displays the minimum and maximum values across all filtering conditions.

Filter	Knee Power (W)		Ankle Power (W)	
	Pre	Post	Pre	Post
6Hz	30.1 (15.2)	30.2 (17.5)	173.9 (57.4)	183.6 (48.9)
10Hz	28.5 (14.7)	29.2 (17.3)	176.1 (59.0)	188.4 (55.8)
15Hz	28.8 (15.0)	29.3 (17.1)	177.9 (59.4)	190.6 (57.8)
20Hz	29.2 (15.2)	29.6 (17.1)	177.4 (59.2)	191.0 (58.2)
25Hz	32.9 (17.0)	32.0 (17.9)	178.2 (59.1)	189.5 (55.6)
M6F10	28.5 (14.8)	28.4 (16.7)	182.5 (60.2)	192.6 (50.8)
M6F15	28.5 (14.8)	28.3 (16.6)	185.3 (60.9)	195.4 (51.0)
M6F20	28.5 (14.8)	28.2 (16.6)	186.1 (60.9)	196.0 (51.0)
M6F25	28.5 (14.8)	28.2 (16.6)	186.2 (60.9)	196.1 (50.9)
M10F15	28.5 (14.7)	29.0 (17.3)	179.2 (59.5)	191.7 (55.9)
M10F20	28.5 (14.7)	28.9 (17.3)	180.1 (59.5)	192.5 (55.7)
M10F25	28.5 (14.7)	28.9 (17.3)	180.3 (59.6)	192.8 (55.6)
M15F20	28.7 (14.9)	29.2 (17.1)	178.7 (59.5)	191.6 (57.4)
M15F25	28.7 (14.9)	29.2 (17.1)	179.0 (59.5)	191.9 (57.4)
M20F25	29.2 (15.2)	29.6 (17.0)	177.6 (59.2)	191.4 (58.1)
Raw	28.5 (14.9)	28.1 (16.5)	186.8 (61.7)	196.0 (50.9)
Range	28.5-32.9	28.2-32.0	173.9-186.8	183.6-196.0

Table 6.4 pre-and post-kneeling means across participants for the 6 different filtering conditions used in the two-way repeated measures ANOVA for knee flexion at heel strike (HS), knee flexion range of motion (ROM) and vertical loading rate (VLR). The last row ("Range") displays the minimum and maximum values across all filtering conditions.

Filter	Knee Flexion at HS (Degrees)		Knee Flexion ROM (Degrees)		VLR (N/s)	
	Pre	Post	Pre	Post	Pre	Post
6Hz	-7.2(5.3)	-8.4(5.0)	38.1(3.9)	38.0(2.4)	45.5(12.7)	44.7 (11.9)
10Hz	-7.9(5.4)	-9.3(5.1)	37.9(3.9)	37.5(2.8)	59.8 (17.2)	58.4 (17.2)
15Hz	-8.3(5.5)	-9.6(5.1)	38.1(4.0)	37.8(2.7)	77.0 (21.8)	74.1 (23.9)
20Hz	-8.4(5.5)	-9.7(5.3)	38.1(4.0)	37.8(2.9)	90.9 (28.4)	87.6 (34.0)
25Hz	-8.4(5.5)	-9.8(5.5)	38.1(4.0)	37.8(2.9)	102.0 (36.0)	98.3 (45.4)
Raw	-7.3(5.5)	-8.7(5.5)	38.5(3.9)	38.1(2.4)	163.4 (84.6)	154.9 (103.8)
Range	(-7.2)- (-8.4)	(-8.4) – (-9.8)	37.9 - 38.5	37.5 – 38.1	45.5– 163.4	44.7– 154.9

Table 6.5 A table containing the post-hoc analysis using Tukey Honest Significant Difference on the main effect of filtering condition for the peak vertical loading rate. Rows highlighted in grey indicate a significantly different pair of conditions in the post-hoc test. The lower limit represents the lower 95% confidence interval where the upper limit represents the upper 95% confidence interval

Group A	Group B	Lower Limit	A-B	Upper Limit	P-value
6Hz	10Hz	-48.9	-4.0	20.9	0.8634
6Hz	15Hz	-65.4	-30.4	4.4	0.1274
6Hz	20Hz	-79.0	-44.1	-9.2	0.0043
6Hz	25Hz	-89.9	-55.1	-20.2	0.0001
6Hz	RAW	-148.9	-114.0	-79.1	0.0000
10Hz	15Hz	-51.4	-16.5	18.4	0.7594
10Hz	20Hz	-65.0	-30.1	4.8	0.1366
10Hz	25Hz	-76.0	-41.1	-6.2	0.0104
10Hz	RAW	-134.9	-100.0	-65.1	0.0000
15Hz	20Hz	-48.5	-13.6	21.3	0.8763
15Hz	25Hz	-59.5	-24.6	10.3	0.3383
15Hz	RAW	-118.4	-83.6	-48.7	0.0000
20Hz	25Hz	-45.8	-11.0	24.0	0.9481
20Hz	RAW	-104.8	-69.9	-35.0	0.0000
25Hz	RAW	-93.9	-59.0	-24.1	0.0000

7 Discussion

Recall that the primary purpose of this study is to examine the effects of an acute kneeling exposure on gait parameters of young healthy individuals. It was hypothesised that, following kneeling, there would be:

1. an increase in knee flexion angle at heel strike and greater knee flexion range of motion during stance
2. an increase in peak knee adduction moment and decrease in peak knee flexion moment during the first 50% of stance
3. a reduction in power generation at the knee (second positive peak knee power) and greater power generation at the ankle (peak positive ankle power) during the last 50% of stance
4. a change in the peak vertical loading rate in the first 10% of stance

7.1 Gait Analysis and Kneeling Exposure

The results of the current study suggest some evidence of changes in gait characteristics resulting from an acute exposure to static kneeling. The acute changes in knee flexion angle at heel strike, knee flexion moment and peak vertical ground reaction force indicate an altered ambulatory pattern, which is believed to be the one of the potential causes of knee OA initiation (Edd et al., 2018; Kajaks et al., 2015). The first hypothesis of our primary objective is partially supported where the 30-minutes of kneeling elicited an increase in knee flexion angle at heel strike but did not affect the knee flexion range of motion. The second hypothesis is rejected as there was no significant change in KAM between pre- and post-kneeling, and the change in KFM occurred in the opposite direction than what was predicted. The third hypothesis was also rejected as there was not a significant difference in knee or ankle power generation. The fourth and final hypothesis was rejected as we did not observe a change in peak vertical loading rate in the first 10% of stance. It is possible that these characteristics of gait, which we hypothesized would result from the acute kneeling exposure but did not, may be a chronic coping mechanism for individuals with OA rather than a strategy to alleviate acute changes following a single kneeling exposure.

7.1.1 Knee Flexion Angle

The mean difference in knee flexion angle in our study is relatively small (an increase of 1.6 degrees flexion) and should be interpreted with caution. The minimal detectible difference (MDD) calculated from each participant's pre-kneeling gait trials was 6.0 degrees. This is similar to the MDD of 5.6 degrees previously reported on young healthy participants (Wilken et al., 2012). Earlier work on individuals who had experienced knee straining postures as part of their occupation for the past 5 years found that this population had measured knee flexion angles at heel strike of 17.1 (+/- 9.6) degrees compared to the matched controls who had on average 10.3(+/- 5.6) degrees of knee flexion at heel strike (Gaudreault et al., 2013). That's an approximate 6.8-degree increase in knee flexion angle at heel strike for knee straining workers compared to non-knee straining controls. That increase is above the conservative 6.0 degree cut-off for meaningful change that was calculated from the current data set and others (Wilken et al., 2012). It is reasonable that an acute exposure did not approach this magnitude of change, however since it is statistically significant it points to the idea that with chronic exposure comes larger/compounding changes in gait.

The previous study on knee straining workers (Gaudreault et al., 2013) included 18 healthy workers with no diagnosis of clinical or radiological knee OA who were exposed to occupational knee straining postures defined as sustained deep knee flexion (kneeling or squatting) during work hours for at least 30min/day for the past 5 years. This knee straining group was compared to a matched control group of

20 participants who have not been exposed to occupational knee straining postures. The gait kinematics were recorded using an electromagnetic motion tracking system, while participants walked on a treadmill at their preferred speed. Unlike our study that only compared knee flexion angle at heel strike and range of motion over stance phase, Gaudreault and team compared knee flexion/extension, abduction/adduction and internal/external rotation range, angle at foot contact, peak angle and minimum angle during swing and stance phase. They found that knee straining workers had greater knee flexion at heel strike, lower peak flexion angle, and lower angle range than non-knee straining workers (Gaudreault et al., 2013). In addition, all measurements taken in the frontal plane showed that knee straining workers had their knees more adducted than non-knee straining workers (Gaudreault et al., 2013). So, although the mean difference we found was relatively low, it is important to acknowledge we found a difference in gait after one 30-minute exposure in the same direction as the change seen after 5 years of daily 30-minute exposures. It is reasonable to speculate, with this evidence, that with greater exposure comes greater change.

Previous work on cartilage thickness and gait kinematics has found that the location of the thickest cartilage on the femur and tibia were largely correlated to the knee flexion angle at heel strike (Koo et al., 2011; Scanlan et al., 2013). In the current study, the small increase in knee flexion angle at heel strike means that the knee is less extended when the foot first contacts the floor. It is possible that individuals who work in a setting where they are exposed to knee straining postures such as squatting and kneeling and lack full knee extension at heel strike load different regions of cartilage during gait (Gaudreault et al., 2013; Scanlan et al., 2013). As gait parameters change with kneeling exposures, the location of cartilage on the femur and tibia that receives loading during gait would also change and would be required to adapt (Edd et al., 2018).

7.1.2 Knee Moments

The increase in peak knee flexion moment was not expected as this was in the opposite direction of what is typically seen in individuals with knee OA. The increase in external knee flexion moment is balanced by the internal knee extensor muscle moment, suggesting that following kneeling, participants are placing greater demands on their quadriceps which is contrary to the compensations adopted by most individuals with knee OA. Some previous meta-analyses have reported conflicting evidence for the association with peak knee flexion moment with knee OA (Mills et al., 2013; Van Tunen et al., 2018). Initially, the increase in knee flexion angle and peak knee flexion moment could be interpreted as participants could be aiming to reduce joint rate of loading by accepting their weight with more knee flexion. In contrast, weight acceptance with more knee extension, typical of osteoarthritic gait, is related to greater axial loading rate at the knee. However, as discussed earlier there was no significant difference in the peak vertical ground reaction loading rate in the first 10% of stance so this is likely not the explanation for this current data set.

Although there was no significant difference in the pre- and post-kneeling knee adduction moment, there is a visible change where it appears to have increased in response to the kneeling exposure (Appendix Figure D.2). The study done by Wilken et al. (2012)reports that within-subject and between-group differences less than 0.11 Nm/kg and 0.13 Nm/kg for peak knee flexion moment and knee adduction moment respectively should be interpreted with caution. To get a rough estimation of how our results compare, the mean difference in peak knee flexion moment and knee adduction moment for all 14 subjects was 3.87 Nm and 1.40 Nm which are also below the calculated MDD from the pre-kneeling trials of 15.38 Nm and 7.31 Nm respectively (Table 6.1). If we normalize the mean difference by the average body mass of participants (70.8kg) we get an estimate of 0.06 Nm/kg and 0.02 Nm/kg which is well below the suggested minimal detectable difference (Wilken et al., 2012). However It should also be noted that a similar study done on 40 healthy young adults using a slightly different 30-minute kneeling protocol where individuals repeated cycles of two minutes of single arm supported kneeling followed by 30 seconds of full-flexion plantarflexed kneeling and a 30 second standing break after every 10 minutes did find that immediately after the kneeling exposure the peak KAM was significantly greater (Tennant et al., 2018). The mean difference detected in that study was 0.18 Nm/kg which is greater than the suggested 0.13 Nm/kg minimal detectable change. Interestingly 30 minutes after exposure the peak KAM was no longer different from baseline (Tennant et al., 2018). So, although it is possible that our study is under-powered, significant and meaningful changes in knee moments following 30 minutes of kneeling have been reported (Tennant et al., 2018). It is also plausible that, like the small change in knee flexion angle at heel strike, the changes in these variables after a single bout of kneeling are not large enough to be detectable or meaningful but could indicate the potential for meaningful change after long-term exposure.

A previous study using a the same kneeling protocol reported in this thesis found a significant increase in variation for peak knee flexion moment and peak knee adduction moment in addition to knee flexion angle and adduction angle post-kneeling (Kajaks & Costigan, 2015). From these results and in support of their model (Figure 2.2), they concluded that the increase in subject waveform variance following kneeling implies that the kneeling protocol caused changes severe enough to acutely alter the loading patterns at the knee joint. The differences we found, although small, support this claim, and suggest a potential mechanism in which kneeling and high knee flexion are a risk factor for knee OA based on the duration of the effects of prolonged static kneeling or cumulative kneeling (Kajaks & Costigan, 2015).

7.1.3 Knee and Ankle Power

A reduction in the second peak knee power is assumed to be a by-product of the "quadriceps avoidance gait" or "stiff gait" that has been observed in individuals with knee OA. Our participants did not adopt a stiff gait as we expected, however this is the first-time mechanical power of the lower limbs has been studied with a kneeling exposure. Previous studies on lower limb mechanical power compared severe OA participants to healthy elders (McGibbon & Krebs, 2004; Segal et al., 2009). It should be acknowledged that one of the most obvious compensations used by individuals with OA is walking slower, which would inherently reduce all joint powers especially ankle power at push-off which is a general characteristic of disablement and aging (McGibbon et al., 2002; Resende et al., Kirkwood, 2012). Another factor in the knee and ankle power compensations that have been observed in individuals with knee OA is the adoption of "stiff" gait and less knee ROM. For example, prior to push-off the knee normally undergoes a second positive peak where the knee extends with an internal knee extension moment. At this time the energy is transferred distally and may contribute to the energy that is ultimately delivered to the foot at push-off. McGibbon and colleagues (2002) found that the lack of the second peak in individuals with OA was related to reaching full knee extension earlier, causing a drop in angular velocity and only allowing minimal energy transfer. Generally, our participants were more flexed at the knee throughout stance phase following kneeling, with no significant difference in knee flexion ROM, so it seems logical that they were able to maintain similar angular velocities unlike individuals with knee OA.

7.1.4 Vertical Loading Rate

We did not find significant differences in the peak vertical loading rate between pre- and post-kneeling. The study done by Tennant et al. (2018) also reported no significant difference in peak vertical rate of loading following a 30-minute, dynamic kneeling protocol. It has previously been published that an increase in vertical loading rate is associated with an increase in axial rate of loading on ankle, knee and hip joints (Mündermann et al., 2005). If this finding remained consistent for long-term occupational high flexion exposures, an increase in axial rate of loading may not be the driver of change in cartilage loading following kneeling. Instead, it is possible the location and magnitude of the loading on the cartilage has changed following kneeling based on the change in knee flexion angle during weight acceptance that was found in this study and others (Gaudreault et al., 2013).

It should also be noted that peak vertical ground reaction forces have been reported to change with gait speed and with knee flexion restriction (Cook, 1997). We were not able to measure gait speed or step length due to limitations in this data set, including tracking only the right lower limb. However, stance time was calculated post-hoc and the mean stance time was 0.70s for both pre-and post-kneeling gait trials. A post-hoc paired t-test was also conducted and produced insignificant results (Appendix E) so it is unlikely that this change in peak vertical ground reaction force is caused by a change in gait speed.

7.1.5 Exploratory sex comparison

After visual inspection of the subject means pre- and post-kneeling, it also appears that there were individuals with different types of responses (Figures 6.1-6.5). Previous reports on acute kneeling and gait adaptations of individuals with knee OA have presented findings where males and females compensate differently (Segal et al., 2009; Tennant et al., 2018). Acknowledging that the sample size in this study was too small for sex comparisons, a preliminary post-hoc analysis was done to explore the possible sex differences on normalized data to account for any differences in body size (Moisio et al., 2003). Table **7.1** presents the normalized mean difference and the mean differences for the male and female participant groups. Of particular interest are the highlighted variables of flexion range of motion, peak ankle power, and peak vertical loading rate, where it appears that, on average, males and females compensated in different directions. Previous work has shown that males and females increase ankle power early on after the onset of knee OA, and this changes as the disease progresses where males decrease ankle power during late stance (Segal et al., 2009). However, to my knowledge there have not been established sex differences in the other variables. It is possible that since the mean differences are in the opposing direction, having the sexes grouped together in the current analysis may have resulted in larger confidence intervals and hindered our ability to detect any actual differences. A larger sample is required to evaluate differences in gait strategies after kneeling and its relation to sex.

Table 7.1 Mean differences from pre-and post-kneeling gait trials divided into male and female participant groups.

Variable	Mean Difference (n=14)	Males (n=8)	Females (n=6)
Heel Strike (degrees)	-1.6	-1.6	-1.5
Flexion ROM (degrees)	-0.1	-0.5	0.5
Knee Flexion Moment Peak (%BW*Ht)	0.40	0.34	0.47
Knee Adduction Moment Peak (%BW*Ht)	0.16	0.19	0.12
Second Peak Positive Knee Power(W/kg)	0.00	0.01	-0.02
Peak Ankle Power (W/kg)	0.22	-0.14	0.71
Peak Vertical Loading Rate (%BW/s)	0.06	0.22	-0.15

7.1.6 Summary

The collective findings from this project could support the mechanical pathway of OA development related to kneeling exposure, where, following a kneeling exposure as short as 30 minutes, some individuals adopt a different gait, potentially to reduce the articular forces during walking and reduce discomfort (Edd et al., 2018; Kajaks & Costigan, 2015). It's possible that, following kneeling, the increase in the loading environment represented by the significant increase in peak knee flexion moment could be a predecessor to knee OA by affecting the homeostatic relationship of the knee with continuing exposure (Edd et al., 2018). The increase in knee flexion angle at heel strike could be a compensatory mechanism to dampen the impact of the increased loading environment at the knee. In young healthy individuals this is likely to be adaptive where the increase in load, with the opportunity to recover, could increase the cartilage thickness and bone density on the tibial plateau (Edd et al., 2018).

There were two potential mechanism of gait changes following kneeling proposed by Kajaks & Costigan (2015)(Figure 2.2). Firstly, kneeling may induce ligament creep, potentially in the medial collateral ligament and posterior cruciate ligament of the knee, due to the prolonged exposure at joint end range, where the creep would then cause knee joint laxity and the change in gait would be a result for compensation of knee instability. Secondly, kneeling could induce some sort of proprioceptive change where the ambulation following includes altered or atypical neuromuscular patterns (Kajaks & Costigan, 2015). The frontal plane laxity analysis that was the original focus of this data set, performed on individuals from the same participant pool, produced insignificant results (Mines, 2016), meaning that there was no significant change in frontal plane laxity after the kneeling protocol. Therefore, it is likely that frontal plane laxity as a proxy for knee joint instability was not the driver of our participants' changes in gait. This leaves potential neuromuscular modifications or changes in proprioception acuity, which were not measured in our study. However work done by Tennant (2016) suggests that following kneeling there was no change in proprioception (characterized by position sense where the individual was asked to reproduce a knee flexion angle of 20 degrees), in young healthy individuals. The flexion angle was selected to evaluate joint position sense at knee flexion angle that approached what was seen at initial contact of gait (Tennant, 2016). Even though no change in proprioception was found, there was still a significant difference in knee adduction moment following the kneeling exposure (Tennant et al., 2018). A delay in muscle onset activation of the vastus medialis muscles was also documented that remained present 30 minutes after the kneeling exposure. This finding points to the idea that following kneeling the alterations in gait could be more related to neuromuscular modifications over changes in knee joint laxity or changes in proprioception (Kajaks & Costigan, 2015b; Mines, 2016; Tennant, 2016; Tennant et al., 2018).

Our study was only intended to show changes in gait immediately following 30-minutes of sustained kneeling. The work by Gaudreault et al. provides evidence that these adaptations could be chronic with repeated exposure to knee-straining activities (Gaudreault et al., 2013). And although we did not directly measure structural changes in the tissues around the knee, it is reasonable to think that, with repeated used of this unnatural gait pattern, it is possible that individuals would then differentially load the cartilage in the knee, potentially leading to pathology (Edd et al., 2018; Kajaks & Costigan, 2015;

Pejhan et al., 2020). For long-term effects on the cartilage to occur, continuous abnormal loading must occur. More work on the dose-response relationship of these changes is needed through longitudinal studies on gait mechanics and individuals exposed to frequent and sustained kneeling. In addition to accounting for covariates of knee OA risk such as obesity, gender, lower limb alignment, and quadriceps strength, and how that might affect the gait compensations following a kneeling exposure.

7.2 Gait Analysis and Filtering Conditions

The secondary purpose of this study was to determine the effects of various low-pass filter cut-offs on knee flexion angles, external knee joint moments, knee joint power, ground reaction forces and loading rates. It was hypothesised that differences in cut-off frequency and large differences in the filter cut-offs between marker and GRF data would result in significantly different knee flexion at heel strike, knee flexion range of motion, peak knee adduction moment, peak knee flexion moment, peak positive ankle power, second peak positive knee power, and peak vertical loading rate.

This work shows that post-kneeling changes in outcomes commonly used in the osteoarthritis literature are robust to changing filter cut-off frequencies apart from peak vertical loading rate. This is likely due to the rest of the measures being far enough away from heel strike, that the additional noise from oscillations caused by the impact at heel strike like that seen during running had minimal effect on these peaks. This finding may suggest that the wide variety of cut-off frequencies (when reported at all) in the literature on gait characteristics in osteoarthritis may not be of significant concern when comparing these outcome measures between studies. However, caution should be taken when using discrete measures in the first 10% of stance such as vertical loading rate.

From visual inspection of a single participant pre-kneeling example (Appendix Figure F.1**)** it appears that the wave forms that used higher frequency cut-offs and mixtures off cut-offs do possess some of the filtering artifact/noise and it is suggested to use those filtering combinations with caution in time-series waveform analyses such as a PCA where the artifact around heel strike may show differences that are not true to human movement.

The scientific process depends largely on replicating and comparing results. For a finding to be considered robust, it must be replicated in multiple studies. Previously published meta-analysis on OA gait characteristics and the progression of OA have produced conflicting evidence that commonly used measurements like knee flexion moments and knee adduction moments are associated with the progression of knee OA. Following our study, it is reasonable to speculate that the difference in published peak lateral and vertical ground reaction forces, knee flexion angle, knee adduction moments, knee flexion moments, knee and ankle power are not due to signal processing differences, but possibly differences in other methodology, and confounding factors.

8 *Limitations*

Some study limitations should be acknowledged. It is likely that some of the dependent variables used in this analysis were under-powered with only 14 participants. Since this was a secondary analysis the number of participants recruited was largely out of the control of the author. The confidence intervals are presented in Appendix Figure G.1. Where most of our results do not provide sufficient evidence to reject the null hypothesis, there appears to be some promise in using this data for future research. Some of the variances are quite large. If the sample size were large enough to control for known covariates of OA such as BMI, sex, quadriceps strength, the real difference between pre-and post-kneeling may be as great as hypothesised. In addition, the population used in our study is considerably younger than age- matched control groups used in most OA and pathological gait studies and, as previously stated, the differences we found were smaller than the minimal detectible change calculated from our own data and those suggested by Wilken et al., 2012. Although it is likely unrealistic to expect to detect a difference as large as that comparing OA populations to controls, or knee straining workers to controls, it seems promising that following just 30 minutes of exposure, we are detecting statistically significant changes that could potentially be compounded over the course of a workday, week, or lifetime.

Gait speed is a confounding factor that we could not account for. Due to the nature of the data, the only spatiotemporal measure we could calculate was stance time, which was calculated post-hoc to confirm that the increase in peak vertical ground reaction force was not influenced by a change in gait speed. The mean stance time was 0.70 seconds for both the pre- and post-kneeling gait trials (Appendix E). A post-hoc paired t-test was also done on participants mean stance time and the results were insignificant. So, it is unlikely that speed influenced our results.

During the data collections, only one sampling rate was used for each system during the data collections. While this is common practice, it should be noted that the filter coefficients depend on that sampling rate (Appendix B). Therefore, theoretically, if the same data could be collected at a different sampling frequency, there would be slightly different filtering outcomes. The participants in this study were barefoot. It is acknowledged that footwear affects walking mechanics, thus those potential effects were controlled by having participants ambulate barefoot. While walking, the markers were strongly secured to the skin, using both tape and straps. While it is not possible to say with absolute certainty that there was no movement of the markers on the skin, there was nothing observed during the data collections, nor in the data to induce suspicion that this might have occurred.

Finally, although we can infer internal loading at the knee, due to the nature of motion capture and the measures used it is not possible at this time to confirm how the internal structures of the knee are being loaded or responding to the kneeling condition and subsequent gait. Given these limitations, is it still important to continue this line of research. It would be interesting to observe possible changes in the magnitude of gait characteristics when accounting for different covariates such as BMI, quadriceps strength, sex, and limb alignment.

9 *Conclusions and Contributions*

This work shows prolonged kneeling has the potential to acutely alter the subsequent knee loading patterns experienced. The findings support the possibility that altered gait because of kneeling exposure could be a contributing factor to osteoarthritis development with repeated exposure. It is possible that the associated changes following kneeling can be compounded over time thus resulting in greater variations in ambulation. This study also supports the idea that changes in ambulation following kneeling are likely related to neuromuscular changes rather than changes in knee laxity as was previously proposed. This study was the first to explore kneeling effects on the less commonly reported variables associated with knee OA such as vertical loading rate, knee power and ankle power in addition to knee adduction moment, knee flexion moment and knee flexion range of motion and knee flexion angle at heel strike. In addition, through a secondary analysis of this data, this has been the first study to compare commonly cited filter cut-off frequencies and their effect on a variety of gait variables, providing the research community some reassurance it is likely that the wide variety of filtering combinations are less of a concern when comparing outcome measures between studies.

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Appendices

Appendix B: Filtering equations

The following equations are from Robertson & Dowling, (2003) and are the default set for the Butterworth filter in Visual 3D (C-Motion Inc., Germantown, MD). This approach corrects for number of passes to maintain the desired cut-off frequency. Below are equations to adjust the Butterworth filter.

$$c_{Bw} = \frac{1}{\sqrt[4]{2^{\frac{1}{n}}-1}}$$

$$f^*_{Bw} = f_{Bw} \times c_{Bw}$$

Where f_{Bw} is the desired cut-off frequency, and f^*_{Bw} is the adjusted cut-off frequency necessary to produce the desired cut-off and *n* is the number of filter passes.

$$\omega^*_c = \tan\left(\frac{\pi f^*_{Bw}}{f_{sr}}\right)$$

w^*_c represents the corrected angular cut-off frequency of the lowpass filter. Where f_{sr} *is the sampling rate in hertz. The equations for* K_1 *and* K_2 *below are used to calculate the low pass filter coefficients* a_0 and b_1 *for the Butterworth filter.*

$$K_1 = \sqrt{2}\omega^*_c \quad K_2 = (\omega^*_c)^2.$$

The low pass coefficients become:

$$a_0 = a_2 = \frac{K_2}{1 + K_1 + K_2};\ a_1 = 2a_0;$$

$$b_1 = 2a_0\left(\frac{1}{K_2}-1\right);\ b_2 = 1-(a_0 + a_1 + a_2 + b_1)$$

The calculated filter coefficients are then used in the following second order recursive filter.

$$y_n = a_0(x_n + 2x_{n-1} + x_{n-2}) + b_1 y_{n-1} + b_2 y_{n-2}$$

Appendix C: ANOVA tables

Appendix Table C.1 the resulting ANOVA table from a two-way repeated measures ANOVA on knee flexion angle at heel strike pre- and post- kneeling, and 6 different filtering conditions. There is a significant main effect of kneeling condition, as expected. No post-hoc was done because the only planned comparisons were for the kneeling*filtering interaction and filtering condition main effects.

Source	Sum Sq.	d.f.	Mean Sq.	F	Prob>F
Filter Condition	14.03	5	2.805	0.1	0.9926
Kneeling Condition	131.65	1	131.647	4.53	0.0348
Filter Condition*Kneeling Condition	0.84	5	0.169	0.01	1
Error	4530.55	156	29.042		
Total	4677.07	167			

Appendix Table C.2 the resulting ANOVA table from a two-way repeated measures ANOVA on knee flexion range of motion pre- and post- kneeling, and 6 different filtering conditions.

Source	Sum Sq.	d.f.	Mean Sq.	F	Prob>F
Filter Condition	5.36	5	1.071	0.09	0.993
Kneeling Condition	3.58	1	3.5758	0.31	0.5755
Filter Condition*Kneeling Condition	0.46	5	0.0915	0.01	1
Error	1771.26	156	11.3543		
Total	1780.65	167			

Appendix Table C.3 the resulting ANOVA table from a two-way repeated measures ANOVA on peak knee adduction moment pre- and post-kneeling for 16 different filtering conditions.

Source	Sum Sq.	d.f.	Mean Sq.	F	Prob>F
Filter Condition	18.8	15	1.256	0.01	1
Kneeling Condition	225.4	1	225.39	2.63	0.1057
Filter Condition*Kneeling Condition	1.1	15	0.071	0	1
Error	35676.5	416	85.761		
Total	35921.8	447			

Appendix Table C.4 the resulting ANOVA table from a two-way repeated measures ANOVA of peak KFM pre- and post-kneeling for 16 different filtering conditions. There is a significant main effect of kneeling condition, as expected. No post-

hoc was done because the only planned comparisons were for the kneeling*filtering interaction and filtering condition main effects.

Source	Sum Sq.	d.f.	Mean Sq.	F	Prob>F
Filter Condition	85.9	15	5.73	0.02	1
Kneeling Condition	1827.7	1	1827.69	6.97	0.0086
Filter Condition*Kneeling Condition	8	15	0.53	0	1
Error	109010.3	416	262.04		
Total	110931.9	447			

Appendix Table C.5 the resulting ANOVA table from a two-way repeated measured ANOVA of second peak knee power pre- and post- kneeling for 16 different filtering conditions

Source	Sum Sq.	d.f.	Mean Sq.	F	Prob>F
Filter Condition	563.1	15	37.543	0.22	0.9993
Kneeling Condition	153.5	1	153.507	0.89	0.3467
Filter Condition*Kneeling Condition	38.3	15	2.55	0.01	1
Error	71953.8	416	172.966		
Total	72708.7	447			

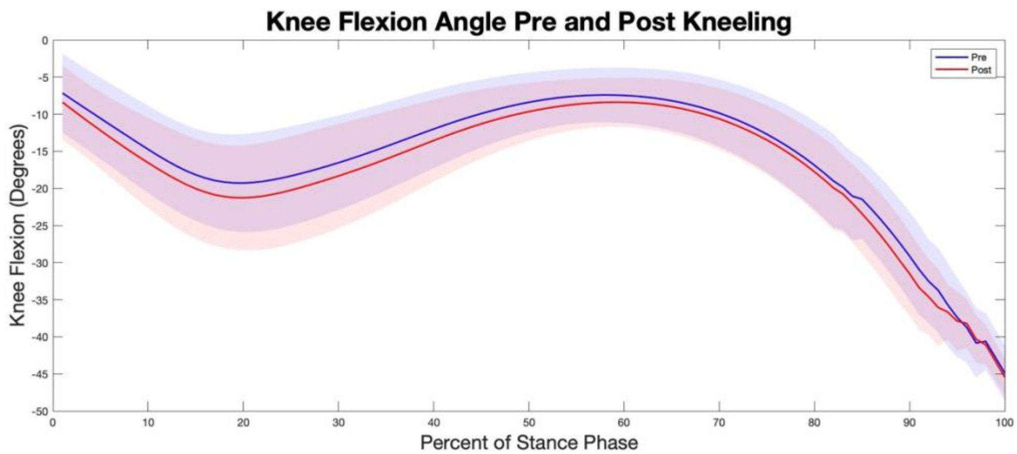
Appendix Table C.6 the resulting ANOVA table from a two-way repeated measured ANOVA of peak ankle power pre- and post- kneeling for 16 different filtering conditions

Source	Sum Sq.	d.f.	Mean Sq.	F	Prob>F
Filter Condition	6200.1	15	413.34	0.21	0.9994
Kneeling Condition	6368.4	1	6368.44	3.28	0.0708
Filter Condition*Kneeling Condition	235.1	15	15.68	0.01	1
Error	807563.4	416	1941.26		
Total	820367	447			

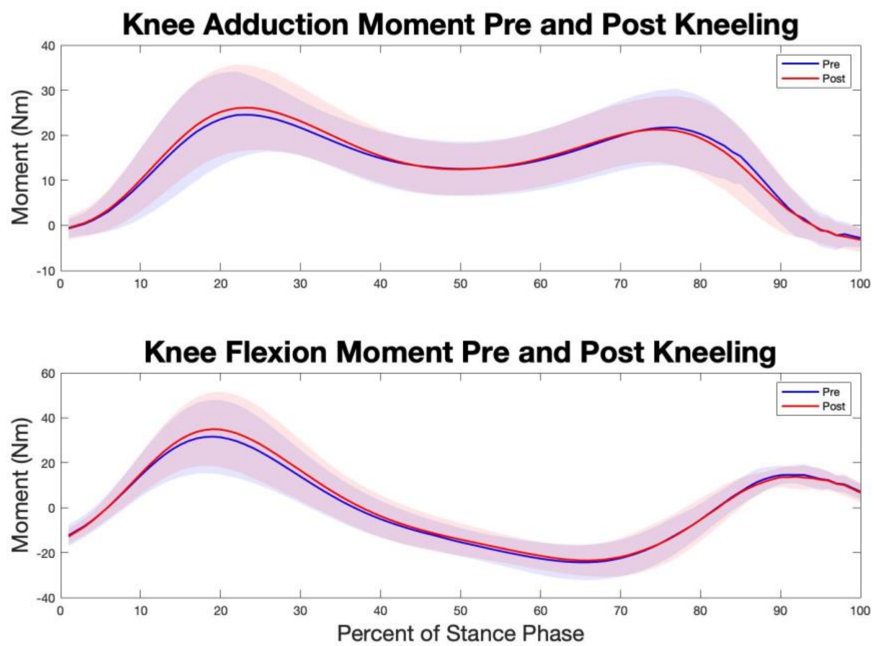
Appendix Table C.7 the resulting ANOVA table from a two-way repeated measured ANOVA of peak vertical loading rate pre- and post- kneeling for 6 different filtering conditions. In this case the 20Hz, 25Hz and Raw filtering condition were significantly greater than the 6Hz condition. The 25Hz and Raw filtering condition were significantly greater than the 10Hz condition and the Raw condition was significantly greater than all other filtering conditions.

Source	Sum Sq.	d.f.	Mean Sq.	F	Prob>F
Filter Condition	285546.8	5	57109.4	36.95	0
Kneeling Condition	1185.8	1	1185.8	0.77	0.3825
Filter Condition*Kneeling Condition	2128.2	5	425.6	0.28	0.9261
Error	241132.4	156	1545.7		
Total	529993.2	167			

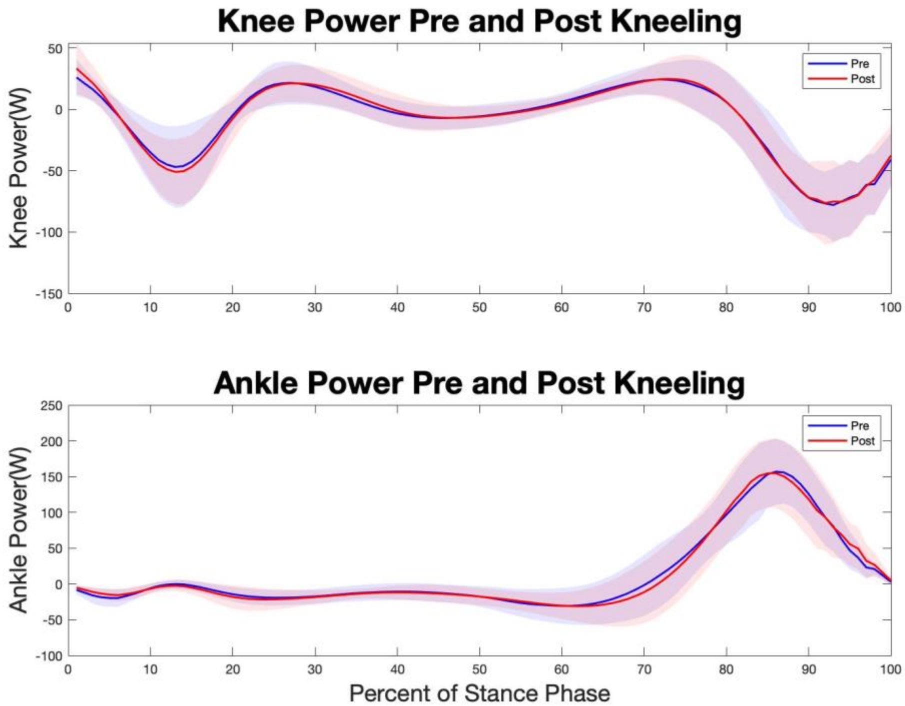
Appendix D : The complete wave forms of the mean subject gait data



Appendix Figure D.1 Mean knee flexion angle for the entire stance phase for all subjects pre- (red) and post-(blue) kneeling gait trials*. The shaded areas represent one standard deviation*



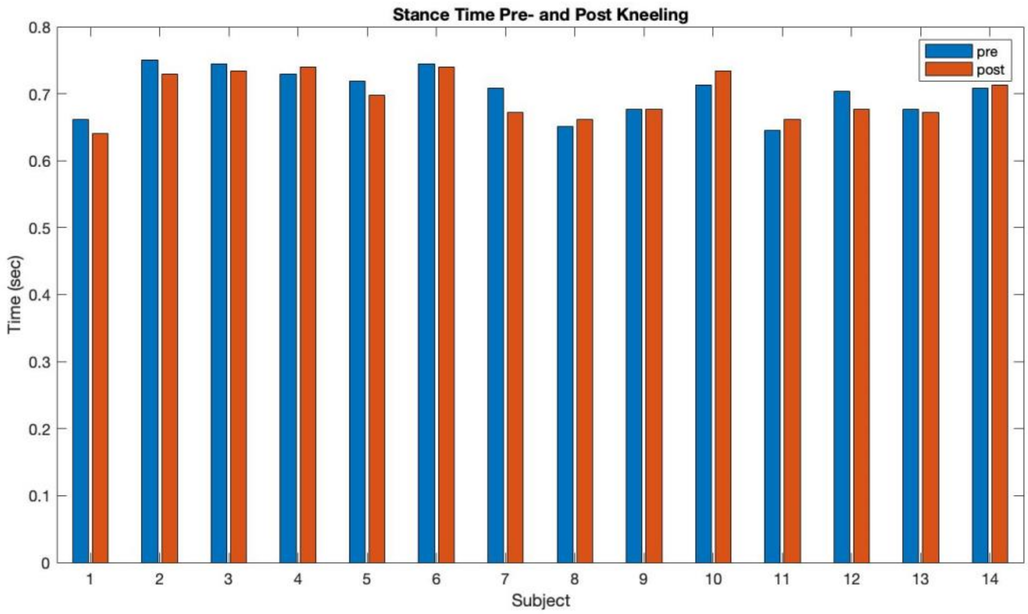
Appendix Figure D.2 Mean external knee flexion moment and mean external knee adduction moment for the entire stance phase for all subjects pre- and post-kneeling gait trials. The shaded areas represent one standard deviation



Appendix Figure D.3 Mean knee power curves and mean ankle power curves for all subjects pre- and post-kneeling gait trials.

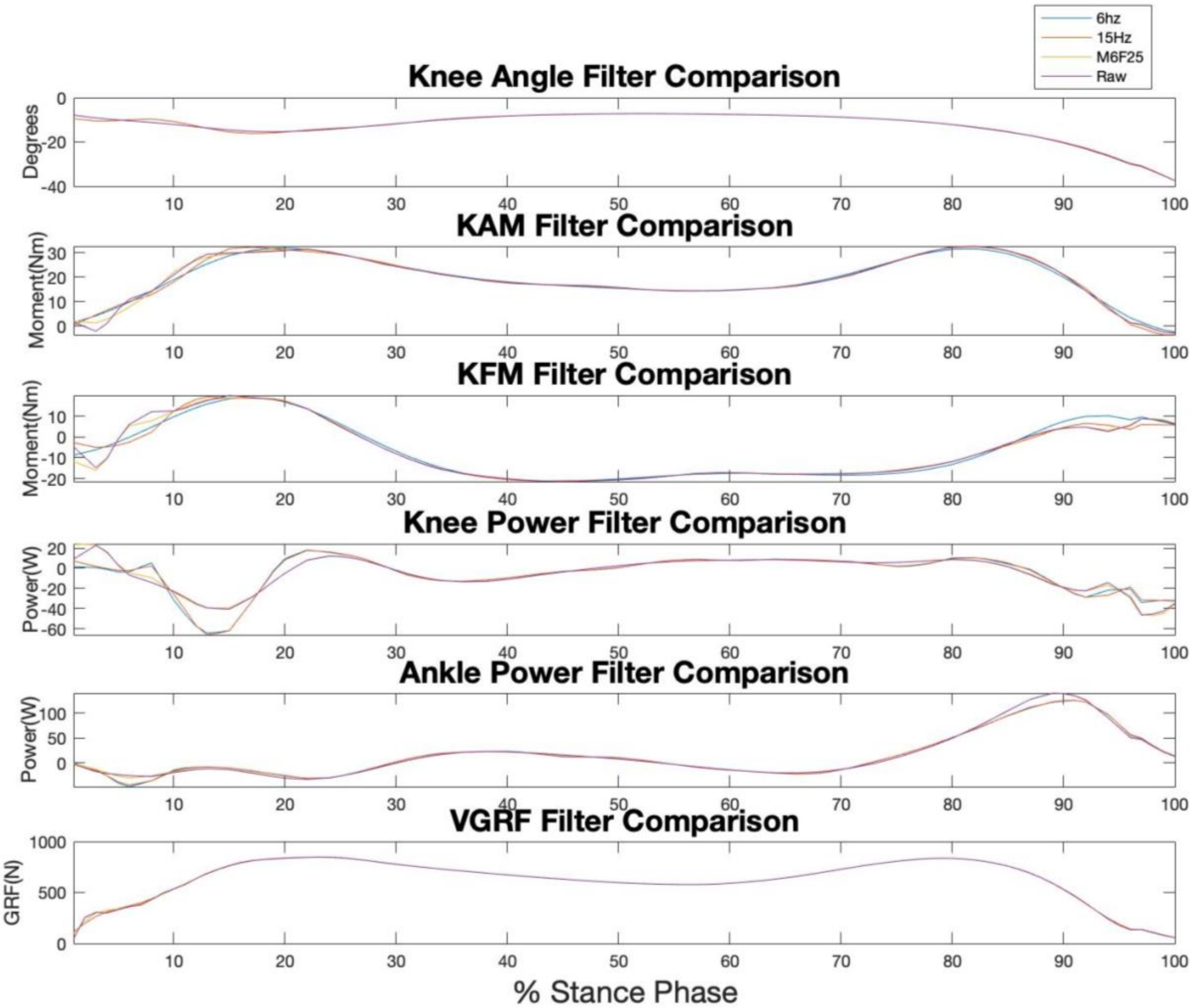
The shaded areas represent one standard deviation.

Appendix E: Post-hoc calculation of stance time



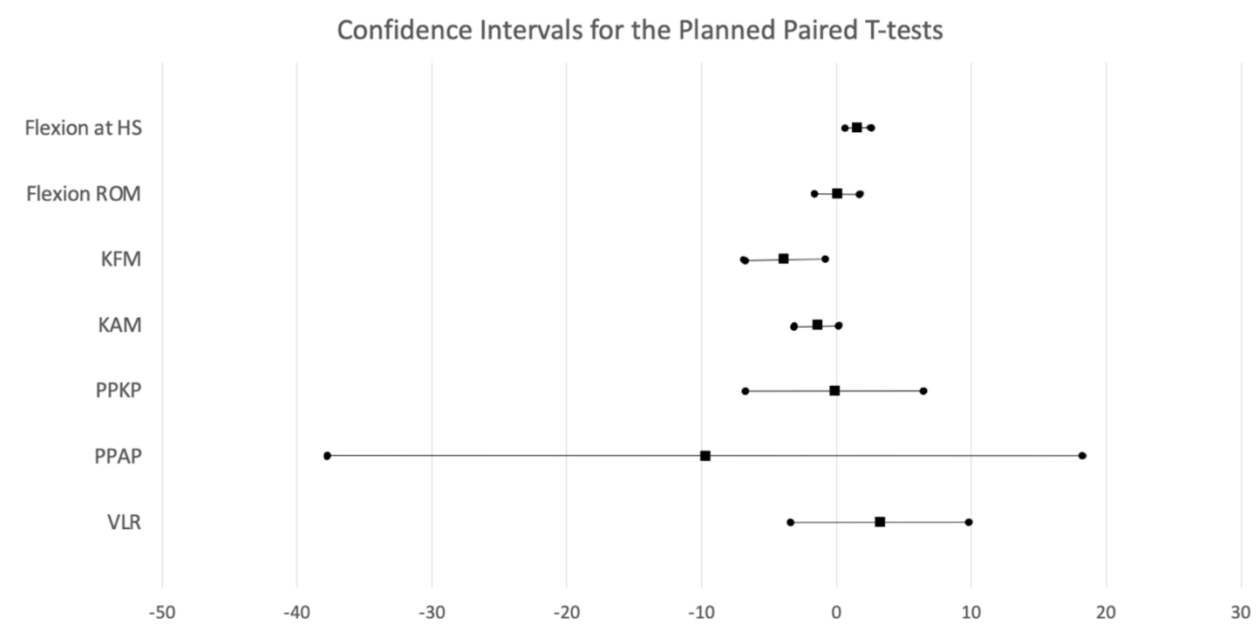
Appendix Figure E.1 A bar graph showcasing the average stance time for each individual pre- and *post-kneeling*. *Mean within participant difference was 0.006s (0.035s).* A post-hoc paired t-test was also conducted between pre- and post-kneeling stance time. Stance time did not significantly change pre- to post-kneeling (t=1.29, df=13, p=0.2208)

Appendix F : Wave Form Comparisons for Filtering Conditions



Appendix Figure F.1 Waveform comparisons of a single participants pre-kneeling gait trial filtered at 6Hz, 15Hz, M6F25 and raw

Appendix G : Confidence Intervals for completed within participant student t-tests (primary objective analysis)



Appendix Figure G.1 Confidence intervals for the 10 planned within participant t-tests that was run in the current study

Appendix Table G.1 A table containing the mean differences and confidence intervals for each within participant students t-test run in this study

Factors	**Mean Difference**	**Upper**	**Lower**
Flexion at HS (degrees)	1.6	2.5	0.6
Flexion ROM (degrees)	0.1	1.8	-1.7
KFM (Nm)	-3.9	-0.9	-6.9
KAM (Nm)	-1.4	0.3	-3.1
PKP (W)	-0.1	6.6	-6.8
PAP (W)	-9.7	18.4	-37.8
VLR (N/s)	3.2	9.9	-3.5