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A Case for Per-Unit-Distance Loads

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ABSTRACT

Peak knee joint contact forces ("loads") in running are much higher than they are in walking, where the peak load has been associated with the initiation and progression of knee osteoarthritis. However, runners do not have an especially high risk of osteoarthritis compared to non-runners. This paradox suggests that running somehow blunts the effect of very high peak joint contact forces, perhaps to provide a load per unit distance traveled (PUD) that is relatively low. **Purpose:** To compare peak and PUD knee joint loads between human walking and running. **Methods:** Fourteen healthy adults walked and ran at self-selected speeds. Ground reaction force and motion capture data were measured and combined with inverse dynamics and musculoskeletal modeling to estimate the peak knee joint loads, PUD knee joint loads, and the impulse of the knee joint contact force for each gait with a matched-pair (within-subject) design. **Results:** The peak load was three times higher in running (8.02 vs. 2.72 bodyweights, \( p < 0.001 \)) but the PUD load did not differ between running and walking (0.80 vs. 0.75 bodyweights\(\cdot\)m\(^{-1}\), \( p = 0.098 \)). The impulse of the joint contact force was greater for running than for walking (1.30 vs. 1.04 bodyweights\(\cdot\)s, \( p < 0.001 \)). The peak load increased with increasing running speed, while the PUD load decreased with increasing speed. **Conclusions:** Compared to walking, the relatively short duration of ground contact and relatively long length of strides in running seem to blunt the effect of high peak joint loads, such that the PUD loads are no higher than in walking. Waveform features other than or in addition to the peak value should be considered when studying joint loading and injuries. **Key Words:** WALKING; RUNNING; KNEE JOINT CONTACT FORCE; MUSCULOSKELETAL MODELING
INTRODUCTION

Chronically high peak joint loads at the knee during walking are suspected to play a causal role in the initiation and progression of knee osteoarthritis. By “load” we refer to the structural mechanics definition of an applied force, in this context specifically the contact force between the articulating tibiofemoral surfaces. Notably, surrogate measures indicating a relatively high peak knee joint load at baseline have been associated with the risks of symptom progression (9,38) and, albeit less convincingly, disease initiation (1,33). However, in research published over 30 years ago, an intense 14-week running program did not accelerate disease progression in rabbits with knee osteoarthritis (55). Although some studies have reported a higher risk of osteoarthritis with higher levels of general physical activity or running specifically (13,50), most studies on humans have concluded that long-distance running is not associated with an increased risk of knee osteoarthritis (16,24). This is not to say that runners are less at risk for knee osteoarthritis compared to non-runners. However, they do not appear to be at greater risk.

Peak knee joint loads in running are about three times greater than those in walking (41,45,46). If peak loads during walking indeed play a causal role in the initiation and progression of knee osteoarthritis, we are left with a paradox concerning running: how can peaks on the order of 2-3 times bodyweight be detrimental to the long-term health of the knee joint if peaks in excess of eight times bodyweight are not? In other words, why do runners not have a greater risk of knee OA than non-runners if they frequently experience peak joint loads that are so high?

This question can be addressed from a variety of perspectives in human movement science. From a biomechanics perspective, gait analysis produces a litany of time-varying
signals that must be reduced to discrete data points for formal statistical analysis. In this study, we wondered if the peak joint load in running is not the most relevant reduction choice when seeking to explain why runners are not especially prone to knee osteoarthritis. The loading response of articular cartilage depends not only on the peak load but also on factors such as the loading rate and duration (12), and the magnitude of a load applied to a biological tissue is only one of many important factors that affect whether the tissue ultimately experiences positive or negative remodeling (27). In particular, previous studies have documented the clinical relevance of variables assessing “average” or “cumulative” loading of the knee and other joints (28,34,40,44,53).

In this study, we wondered if the biomechanics of running protect the knee from peak loads that would otherwise be highly injurious if they were encountered during walking. A seemingly reasonable measure of cumulative/average/total joint loading is the impulse, as it considers both the time-varying magnitude and duration of application of the load. The duration of stance (foot-ground contact) is much shorter in running than in walking, meaning that the duration of load application is relatively brief. In addition, the speed of progression and the stride length are longer in running, meaning that the same distance can be covered in less time and fewer loading cycles than in walking. These comparisons suggest that the disparity between joint loads accumulated per unit distance traveled in walking and running should be much smaller than the disparity in the peak joint loads. The energy cost of locomotion is often investigated on a per-unit-distance-traveled basis, i.e. the metabolic cost of transport, which has provided a number of insightful comparisons between walking and running (10,35,51), but to our knowledge, joint loading has not been assessed on this basis.
Therefore, the purpose of this study was to compare peak and “per-unit-distance” (PUD) knee joint loads between walking and running. We hypothesized that peak joint loads would be greater in running than in walking, but that the PUD load and the impulse would not differ between the two gaits.

METHODS

Subjects

Pilot data suggested approximately 10 subjects were needed for differences in PUD loads equating to “moderate-to-large” effect sizes (15) to reach statistical significance ($\alpha = 0.05$, $\beta = 0.80$). Fourteen healthy adults (male/female 7/7, age 25±11 years, height 1.73±0.11 m, mass 72.4±13.6 kg) participated in this study. Subjects were in good health, were recreationally active, and were excluded if they had a history of major knee trauma or any events in the past year that adversely affected their gait. Most subjects ran for fitness but none were high-level competitive runners. All subjects gave written informed consent prior to participating, and all protocols were approved by the local institutional research ethics board.

Experimental Setup

Motion capture was performed along a level 15-m walkway surrounded by 14 optical motion capture cameras (Oqus 400, Qualisys, Gothenburg, Sweden). The middle 2.4 m of the walkway was instrumented with four tandem 61x61-cm strain gauge force platforms (custom BP model, AMTI, Watertown, MA, USA). Prior to data collection, the force platforms were aligned and calibrated using a CalTester wand (C-Motion, Germantown, MD, USA).
During the experiment, subjects were barefoot and wore a form-fitting spandex shirt and shorts. The set of passive retroreflective markers shown in Fig. 1 was attached to the subject using elastic wraps and double-sided tape.

Protocol

Subjects first performed a calibration trial, standing upright on both feet with the feet shoulder-width apart. Subjects were next instructed to walk and run along the walkway at self-selected “normal and comfortable” speeds. The walking speed was instructed to be one that the subject would use for “walking down the street”. The running speed was instructed to be one that the subject would use to “jog for exercise”. The motion capture system sampled marker positions at 200 Hz while the force platforms sampled ground reaction forces (GRF) synchronously from the force platforms at 1000 Hz. Subjects performed five trials of each gait, with at least one full stride (heel-strike to heel-strike) of the right leg measured per trial. Trials were performed in blocks. The order of the walking and running conditions was randomized.

Data Analysis

Marker coordinates were digitized and labeled in Track Manager (Qualisys, Gothenburg, Sweden) and exported to Visual3D (C-Motion, Germantown, MD, USA) for further analysis. Marker positions were smoothed using a fourth-order recursive lowpass Butterworth filter with cutoff frequencies of 6 Hz for walking and 10 Hz for running. The GRF were smoothed at 75 Hz. A rigid body linked-segment model was defined for each subject from the average marker positions in the standing calibration trial. The model considered the full body (Fig. 1) but only the pelvis and lower limb data were used in this study. An orthogonal local coordinate system
was defined for each segment (see Appendix A, SDC 1, details on kinematic modeling). Three-
dimensional joint angular positions were calculated from the tracking markers using the 6DOF
method (11). Joint moments of force were calculated by inverse dynamics (see Appendix B,
SDC 2, details on kinetic modeling).

Next, a musculoskeletal modeling analysis was performed in Matlab (The MathWorks,
Natick, MA, USA). A generic lower limb musculoskeletal model (3) was scaled to the segment
lengths of each subject and used to calculate the moment arms and lines of action for 44 muscles
from the input joint angles. Static optimization was used to calculate muscle forces at each
timestep that minimized the sum of the squared muscle stresses (22) and reproduced each
subject’s hip flexion/extension, hip adduction/abduction, knee flexion/extension, and ankle
dorsi/plantarflexion moments from the inverse dynamics, which were implemented as equality
constraints. The muscles were modeled as ideal force generators with no contractile or elastic
properties (2). Optimizations were performed using the interior-point algorithm in the Matlab
Optimization Toolbox.

The knee joint load (axial knee joint contact force on the tibial plateau) was then
calculated knowing the resultant knee joint force vector from inverse dynamics and the knee
muscle force vectors from static optimization, and was scaled by the subject’s bodyweight (BW).
The impulse of the joint contact force was calculated as the integral of the force-time series:

\[ F_c \Delta t = \int_0^T F_c(t) \, dt \]
where \( F_c \) is the joint contact force. The PUD load (units = BW-m\(^{-1}\)) was calculated as the average contact force during all complete strides (equivalent to a time-normalized impulse), divided by the stride length:

\[
\frac{F_c}{\Delta x} = \frac{\int_0^T F_c(t) \, dt}{T \cdot L}
\]

where \( T \) is the stride duration and \( L \) is the stride length. The PUD load is essentially the average load per stride, accounting for the distance covered per loading cycle, which is typically longer in running than in walking. PUD loads and impulses were then averaged over strides to compute representative values for walking and running for each subject. We also performed the same calculations on the knee adduction moment (KAM), a commonly used surrogate for joint loading in gait analysis (47) that is predictive of OA progression risk (9,38). Moments were expressed in multiples of BW times standing height (BW-Ht).

**Statistical Analysis**

Matched-pair Student’s \( t \)-tests, with each subject serving as their own control, were used to compare the peaks, PUD loads, and impulses between walking and running. A critical value of \( p = 0.05 \) was assumed for significance. To complement these tests, we also calculated effect sizes (Cohen’s \( d \)) to assess the substantive meaningfulness of the differences between walking and running. Using Cohen’s (15) guidelines, values of \( d > 0.2, 0.5, \) and \( 0.8 \) were assumed to indicate small, moderate, and large differences, respectively.
To further explore the relationship between running and the peak and PUD knee joint loads, we reanalyzed knee joint contact force data in 10 subjects at different running speeds (2.5, 3.5, and 4.5 m\( \cdot \)s\(^{-1} \)), calculated using similar models and methods as described here (17). The relationships between increasing running speed and the outcome variables were tested by linear regression.

RESULTS

The walking speeds chosen by the subjects averaged 1.45±0.12 m\( \cdot \)s\(^{-1} \) (range 1.17-1.65 m\( \cdot \)s\(^{-1} \)) while the running speeds averaged 3.17±0.43 m\( \cdot \)s\(^{-1} \) (range 2.32-3.91 m\( \cdot \)s\(^{-1} \)). The average preferred stride lengths were 1.44±0.14 m for walking and 2.26±0.41 m for running. Stance duration averaged 0.61±0.06 s (61.2±1.6% of the stride) for walking and 0.25±0.02 s (35.7±4.4% of the stride) for running. The mean joint angle and joint moment time series for both gaits are presented in Fig. 2.

The peak knee joint contact force was greater (\( p < 0.001 \); Fig. 3a) in running (8.02±1.62 BW) than in walking (2.72±0.41 BW), with a large effect size (\( d = 3.44 \)). However, the PUD load did not differ (\( p = 0.098 \); Fig. 3b) between running (0.80±0.14 BW\( \cdot \)m\(^{-1} \)) and walking (0.75±0.08 BW\( \cdot \)m\(^{-1} \)), and had a small-to-moderate effect size (\( d = 0.48 \)). The peak KAM also did not differ (\( p = 0.142; \ d = 0.31; \) Fig. 4a) between running (3.32±2.26 BW\( \cdot \)Ht) and walking (2.72±0.80 BW\( \cdot \)Ht), but the PUD KAM was smaller (\( p = 0.002; \ d = -1.26; \) Fig. 4b) for running (0.27±0.24 BW\( \cdot \)Ht\( \cdot \)m\(^{-1} \)) than for walking (0.53±0.20 BW\( \cdot \)Ht\( \cdot \)m\(^{-1} \)). The impulse of the joint contact force was greater (\( p < 0.001; \ d = 1.86; \) Fig. 5a) for running (1.30±0.19 BW\( \cdot \)s) than for walking (1.04±0.09 BW\( \cdot \)s), while the KAM impulse was smaller (\( p = 0.008; \ d = -1.14; \) Fig. 5b) for running (0.40±0.32 BW\( \cdot \)Ht\( \cdot \)s) than for walking (0.76±0.32 BW\( \cdot \)Ht\( \cdot \)s).
Linear regression analysis of the Edwards et al. (17) data indicated that peak knee joint contact forces increased with increasing running speed ($p = 0.001$, $r = 0.56$, 31% variance explained), but the PUD load decreased with increasing running speed ($p < 0.001$, $r = -0.77$, 59% variance explained). The impulse of the joint contact force also increased with increasing running speed ($p = 0.01$, $r = 0.46$, 21% variance explained).

DISCUSSION

This study compared peak and PUD joint contact loads at the knee between walking and running. The impulse of the joint load was also assessed. The primary finding was that while the peak joint contact force and the impulse of that force were greater in running, the PUD load did not differ between walking and running. Note that the impulse and the PUD load are not linearly proportional to each other (Fig. 6). Although the difference in PUD loads between walking and running approached statistical significance ($p < 0.10$), the effect size of this difference suggested its magnitude was at most of moderate substantive meaningfulness (18). The study had sufficient power to detect differences as small as 0.072 BW·m$^{-1}$ in the PUD load. The difference required for a large effect size was 0.098 BW·m$^{-1}$, 36% greater than the minimum detectable difference, indicating that the observed difference did not reach statistical or practical significance. We are therefore confident in the conclusion that the PUD load did not differ between walking and running.

The results have implications for the role of running in exercise and joint health. Running is usually considered a “high-impact” activity that can be hard on the knees (14). The present study and others (22,41,46) indicate that peak knee joint loads in running are well above the threshold that appears to promote progression (9,38) and possibly the initiation (1,33) of knee
osteoarthritis in walking. If walking and running shared a common paradigm for joint loading and negative cartilage turnover, we would therefore expect many runners to eventually develop knee osteoarthritis. The present results offer a potential biomechanical explanation for why this expectation is faulty: the duration of load application and the distance covered per loading cycle offset the relatively high peak joint load of running, such that the load accumulated per unit distance traveled is no greater in running than it is in walking. In addition, running has a relatively long swing phase as well as a flight phase, absent from walking, during which joint loads are quite small. Studies on gait modification to treat or prevent knee osteoarthritis may be wise to consider these factors (load duration, stride length), as it seems they are able to blunt the effects of peak joint contact forces that would likely be highly injurious if they were encountered during walking.

This blunting effect was present at slow, moderate, and fast distance running speeds, suggesting that runners may be able to increase their average training speed without increasing their risk of knee osteoarthritis, although this claim would need to be validated longitudinally and it should be considered that the joint contact impulse increased with speed. We also caution that this conclusion likely does not extend to running injuries in general. However, running actually induces positive remodeling of the knee cartilage compared to a sedentary lifestyle (54), suggesting that joint loads from running could be beneficial for preventing knee osteoarthritis. It is currently unknown if this conclusion extends to special at-risk populations (e.g. ACL injuries, overweight, and amputees), or if the PUD load is also similar in walking and running for these populations. These topics may be worthy for consideration in future joint loading studies. Relatedly, it will be important to determine how factors such as cartilage conditioning, knee
alignment, and the mechanical properties of muscle and bone influence an individual runner’s tolerance for high joint loads and the risk of joint degeneration.

The present results also have implications for how knee osteoarthritis and joint loading in general are studied in gait analysis. The effect of mechanical loading on articular cartilage depends on the characteristics of the applied load (12,30,52). For example, cyclical compressive loading improves cartilage stiffness and proteoglycan synthesis, but continuous compressive loading reduces stiffness and impairs synthesis (4). Loss of cartilage stiffness with osteoarthritis is suspected to compromise the ability of the knee cartilage to store and return elastic strain energy during locomotion (48,49). Studies on joint loading often default to using the peak contact force or KAM as the primary outcome variable. However, the peak KAM despite its reputation has actually not consistently distinguished between subjects with and without knee osteoarthritis or different disease severity levels (19,37). Other parameters describing the shape of biomechanical time series data have distinguished between the presence and severity of knee osteoarthritis (5). Maly et al. (34) recently reported that the “cumulative knee adductor load”, defined in their study as the KAM impulse times the number of steps taken in a day, was twice as high in subjects with knee osteoarthritis than in healthy controls, and distinguished between the groups better than the peak KAM. Experiments from sensorimotor control suggest that humans do not always prefer to minimize the peak of a sustained load; a higher peak can actually be preferable depending on the duration and shape of the load (29). We suggest that this paradigm should be considered when making choices on data reduction in joint loading studies. The peak force or moment is unlikely to tell the whole story in a complex loading environment like the knee.
Rat models have induced knee osteoarthritis in previously healthy knees by running (8,42), but the training volumes in these studies (5-10 km/week) were intensive for such a small animal that does not normally engage in prolonged running. More moderate running in rats is beneficial for joint health (21). Extrapolating animal model results to humans is not straightforward, particularly in the case of quadrupeds, but an alternative explanation in light of these results is that most humans spend much less time running than they do walking. Perhaps running simply does not add enough loading cycles to bring the cartilage above an injury threshold. However, we think this explanation is unlikely based on the fatigue response of cartilage to applied loads. Cadaver studies of human articular cartilage have indicated that small changes in the applied stress produce exponentially larger changes in the number of loading cycles until failure (57). If the peak compressive stress on the medial tibiofemoral joint in walking is about 1.4 MPa (7) and if we assume the peak stress in running is about 3x1.4=4.2 MPa, which is almost certainly an underestimate since knee flexion reduces the joint contact area (26), the Weightman (57) model for a young adult predicts that each loading cycle of running induces as much fatigue damage as 34 loading cycles of walking, or that a three-mile run adds as much fatigue stress as 60-70 miles of walking at normal speeds and stride lengths. Consequently, we think that the loading cycles added by even a brief daily run are not negligible, and that factors other than the relatively brief amount of time runners spend actually running must be involved in the lack of a higher prevalence of knee osteoarthritis in runners.

While the peak joint contact force was greater in running than in walking, the peak KAM did not differ between gaits. Similarly, the PUD load did not differ between walking and running, but the PUD KAM and impulse were smaller for running. The KAM in general is undoubtedly a useful outcome variable for studying joint loading, as it is clearly correlated with...
the risk of knee osteoarthritis (1,9,33,38). However, the present findings support the conclusions of Walter et al. (56) and Meyer et al. (36) that changes in the KAM do not necessarily reflect changes in the actual joint contact force. This force cannot be measured in vivo in most cases, which motivates the use of musculoskeletal modeling and muscle force prediction to estimate changes in joint loading that complement assessment of the KAM. Joint load estimation requires a variety of modeling assumptions that are difficult to validate, but the peak joint contact force estimated for walking in this study (2.72 BW on average) was within the range of 2.00-3.25 BW measured in five subjects with instrumented joint replacements (31). The accuracy of these predictions should hopefully improve with further progress on best practices for the associated modeling and simulation methods (20). We note that the KAM and joint kinetics in general from inverse dynamics are also model-based variables, not experimental measurements (25,39). They are surrogate metrics for joint loading, not actual joint loads, and are subject to modeling assumptions and limitations that should be scrutinized and considered in their interpretation. If we assume the model-based predictions of the joint contact forces are reasonably accurate, the present results exemplify how the KAM can potentially give a misleading picture of joint loading. According to the KAM results alone, we would conclude that the peak “load” did not differ between walking and running, and that the PUD load and the impulse were smaller in running. None of those conclusions are consistent with the changes seen in the joint contact force variables.

There are several limitations to the present study that are worthy of comment, although we note that the impact of most of them should be minimized by the within-subjects design. Knee osteoarthritis symptoms most often appear on the medial compartment of the tibiofemoral joint, but the present study estimated the total joint contact force rather than a specific medial
compartment force. To assess this limitation, we calculated medial knee joint contact force using the method described by Winby et al. (58), which produced the same results of statistical significance as our analysis of the total joint contact force. The use of self-selected vs. prescribed speeds is a contentious topic in gait analysis. Our rationale for allowing subjects to select their own walking and running speeds was that we sought to replicate the loading conditions they experience in every-day life, which are presumably more relevant to understanding osteoarthritis development than forcing all subjects to use the same speeds (6). Forcing all subjects to use the same speeds may have reduced the between-subjects variance in the outcome variables, but this variance was not an important element of this study. Running barefoot may have altered subject’s footstrike patterns (23) but the effect on joint loading would be negligible compared to the differences between walking and running overall.

The musculoskeletal model for static optimization requires a large number of input parameters (e.g. muscle physiological cross-sectional areas and musculoskeletal geometry), which were defined using a generic lower limb model based mostly on cadaver studies (3). The use of subject-specific musculoskeletal model parameters, obtained for example by MRI, would likely affect the shape and magnitude of the estimated joint contact forces. However, static optimization results are relatively insensitive to model parameter values compared to their sensitivity to the input joint angles and moments (43). In addition, since the model parameters are the same for both the walking and running conditions, we would not expect subject-specific parameters to affect the statistical significance between these conditions. A related limitation is the use of ideal force generators to model the muscles, but previous studies have suggested this limitation is negligible for muscle force prediction in normal locomotion (2,41). We caution that the extension of these methods to studying osteoarthritis populations would likely require a
modified cost function that considers impaired neuromuscular control, rather than the generic cost function (sum of squared muscle stresses) used here for unimpaired subjects.

In conclusion, running in individuals with healthy knees does not induce PUD joint loads that are any greater than those experienced during walking. This effect is due to the relatively short duration of ground contact, and the relatively long distance covered during a loading cycle. While this result does not imply that runners are at a reduced risk of knee osteoarthritis compared to non-runners, it offers a biomechanical explanation for why running does not seem to increase the risk of osteoarthritis even though peak joint loads in running are very high. Factors other than the peak joint load or peak KAM should be considered when assessing knee osteoarthritis risk and lower limb injury risk in general, e.g. in the context of gait retraining interventions. In particular, measures of the cumulative, average, or total daily load of musculoskeletal structures (34) and analysis of waveform shapes and features (5) seem greatly warranted in future research on walking and running.

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CONFLICT OF INTEREST

The authors have no personal, professional, or financial conflicts of interest related to study. The results of the present study do not constitute endorsement by ACSM.
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FIGURE CAPTIONS

Figure 1. Subject marker locations in the experimental setup. Markers on the head, arms, and trunk were measured but were not used in this study.

Figure 2. Mean joint angles (left) and moments (right) during the stride for walking (solid lines) and running (dashed lines). The stride begins and ends at heel-strike; strides were time-normalized for presentation to facilitate averaging over strides and subjects. Vertical dashed lines indicated toe-off. Shaded areas are +/- one between-subjects standard deviation. Bracketed terms (FL=flexion; AD=adduction; EX=extension; DF=dorsiflexion) indicate the anatomical direction of joint motion for positive (counter-clockwise) values. Moments are presented in multiples of bodyweight (BW) times height (Ht).

Figure 3. Knee joint contact forces during walking (left) and running (right). Error bars are +/- one between-subjects standard deviation. * = significantly greater than walking. (a) Peak loads and (b) PUD loads. Forces are presented in multiples of bodyweight (BW).

Figure 4. Knee adduction moments (KAM) during walking (left) and running (right). Error bars are +/- one between-subjects standard deviation. (a) Peak moments and (b) PUD moments. Moments are presented in multiples of bodyweight (BW) times height (Ht).
Figure 5. Knee joint impulses over the stride cycle for walking (left) and running (right). Error bars are +/- one between-subjects standard deviation. * = significantly different from walking. (a) Impulse of the joint contact force and (b) impulse of knee adduction moment.

Figure 6. Scatter plot of PUD load vs. impulse of the joint contact force for all subjects.

LIST OF SUPPLEMENTAL DIGITAL CONTENT
Appendix A.docx Technical details on the kinematic calculations
Appendix B.docx Technical details on the kinetic calculations
Figure 2

Joint Angles (degrees)

Walking

Running

Joint Moments (BW*Ht)

Walking

Running

Hip FL(+)

Hip AD(+)

Knee EX(+)

Knee AD(+)

Ankle DF(+)

Stride (%)

Stride (%)

Stride (%)
Figure 3
Figure 4
Figure 5

(a) Load Impulse (BW*)

(b) KAM Impulse (BW^*H^*)
APPENDIX A: KINEMATIC MODELING DETAILS

A local orthogonal coordinate system (LCS) was defined for each segment in the standing calibration trial. The pelvis LCS was defined from calibration markers on the left and right anterior-superior iliac spines ($\vec{P}_{ASIS-L}$, $\vec{P}_{ASIS-R}$) and the left and right posterior-superior iliac spines ($\vec{P}_{PSIS-L}$, $\vec{P}_{PSIS-R}$). The pelvis LCS origin ($\vec{O}_{pelvis}$) was located midway between $\vec{P}_{ASIS-L}$ and $\vec{P}_{ASIS-R}$. The pelvis LCS axes were:

\[ \hat{i}' = \frac{\vec{P}_{ASIS-R} - \vec{O}_{pelvis}}{|| \vec{P}_{ASIS-R} - \vec{O}_{pelvis} ||} \]  \hspace{1cm} (Eq. A1)

\[ \hat{v}_{temp} = \frac{\vec{O}_{pelvis} - 0.5 \cdot (\vec{P}_{PSIS-R} + \vec{P}_{PSIS-L})}{|| \vec{O}_{pelvis} - 0.5 \cdot (\vec{P}_{PSIS-R} + \vec{P}_{PSIS-L}) ||} \]  \hspace{1cm} (Eq. A2)

\[ \hat{k}' = \hat{i}' \times \hat{v}_{temp} \]  \hspace{1cm} (Eq. A3)

\[ \hat{j}' = \hat{k}' \times \hat{i}' \]  \hspace{1cm} (Eq. A4)

The thigh LCS was defined from calibration markers on the lateral and medial femoral epicondyles ($\vec{P}_{LFE}$, $\vec{P}_{MFE}$). The thigh LCS origin ($\vec{O}_{thigh}$) was located at the hip joint center. The thigh LCS axes were:

\[ \hat{k} = \frac{\vec{O}_{thigh} - 0.5 \cdot (\vec{P}_{LFE} + \vec{P}_{MFE})}{|| \vec{O}_{thigh} - 0.5 \cdot (\vec{P}_{LFE} + \vec{P}_{MFE}) ||} \]  \hspace{1cm} (Eq. A5)

\[ \hat{v}_{temp} = \frac{\vec{P}_{LFE} - \vec{P}_{MFE}}{|| \vec{P}_{LFE} - \vec{P}_{MFE} ||} \]  \hspace{1cm} (Eq. A6)

\[ \hat{j} = \hat{k} \times \hat{v}_{temp} \]  \hspace{1cm} (Eq. A7)
\[ \hat{i} = \hat{j} \times \hat{k} \]  

(Eq. A8)

Two models of the shank LCS were used. The proximal-biased model defined the shank LCS from calibration markers on the lateral and medial femoral epicondyles and the lateral and medial malleoli (\( \vec{P}_{LMA}, \vec{P}_{MMA} \)). The shank LCS origin (\( \vec{O}_{\text{shank}} \)) was located midway between \( \vec{P}_{LFE} \) and \( \vec{P}_{MFE} \). The proximal-biased shank LCS axes were:

\[ \hat{k} = \vec{O}_{\text{shank}} - 0.5 \cdot \left( \vec{P}_{LMA} + \vec{P}_{MMA} \right) \bigg/ \left| \vec{O}_{\text{shank}} - 0.5 \cdot \left( \vec{P}_{LMA} + \vec{P}_{MMA} \right) \right| \]  

(Eq. A9)

\[ \hat{v}_{\text{temp}} = \frac{\vec{P}_{LFE} - \vec{P}_{MFE}}{\left| \vec{P}_{LFE} - \vec{P}_{MFE} \right|} \]  

(Eq. A10)

\[ \hat{j} = \hat{k} \times \hat{v}_{\text{temp}} \]  

(Eq. A11)

\[ \hat{i} = \hat{j} \times \hat{k} \]  

(Eq. A12)

The distal-biased shank model replaces Eq. A10 and defines the frontal plane with vectors between the malleoli:

\[ \hat{v}_{\text{temp}} = \frac{\vec{P}_{LMA} - \vec{P}_{MMA}}{\left| \vec{P}_{LMA} - \vec{P}_{MMA} \right|} \]  

(Eq. A13)

The distal-biased shank model was used to calculate ankle angles, and the proximal-biased shank model was used to calculate knee angles. The foot LCS was defined from calibration markers on the lateral and medial malleoli, and the first and fifth metatarsal heads (\( \vec{P}_{MET1}, \vec{P}_{MET5} \)). The foot LCS origin (\( \vec{O}_{\text{foot}} \)) was located midway between \( \vec{P}_{LMA} \) and \( \vec{P}_{MMA} \). The foot LCS axes were:
\[
\hat{k}' = \frac{\hat{O}_{\text{foot}} - 0.5 \left( \overrightarrow{p}_{\text{MET}5} + \overrightarrow{p}_{\text{MET}1} \right)}{\left| \hat{O}_{\text{foot}} - 0.5 \left( \overrightarrow{p}_{\text{MET}5} + \overrightarrow{p}_{\text{MET}1} \right) \right|} 
\]  
(Eq. A14)

\[
\hat{v}_{\text{temp}} = \frac{\overrightarrow{p}_{\text{LMA}} - \overrightarrow{p}_{\text{MMA}}}{\left| \overrightarrow{p}_{\text{LMA}} - \overrightarrow{p}_{\text{MMA}} \right|} 
\]  
(Eq. A15)

\[
\hat{j}' = \hat{k}' \times \hat{v}_{\text{temp}} 
\]  
(Eq. A16)

\[
\hat{i}' = \hat{j}' \times \hat{k}' 
\]  
(Eq. A17)

For all segments, the 3x3 rotation matrix \( \mathbf{C} \) was:

\[
\mathbf{C} = \begin{bmatrix} \hat{i}' & \hat{j}' & \hat{k}' \end{bmatrix}^T 
\]  
(Eq. A18)

Joint angles (relative LCS orientations) were extracted from a joint orientation matrix \( \mathbf{R}_{\text{joint}} \) computed at each timestep from the walking and running trials:

\[
\mathbf{R}_{\text{joint}} = \begin{bmatrix} \mathbf{C}_{\text{distal}}^T \mathbf{R}_{\text{distal}} \big| \mathbf{C}_{\text{proximal}}^T \mathbf{R}_{\text{proximal}} \big| \end{bmatrix}^T 
\]  
(Eq. A19)

where \( \mathbf{R}_{\text{distal}} \) and \( \mathbf{R}_{\text{proximal}} \) are the instantaneous rotation matrices of the distal and proximal segments comprising the joint, calculated using a least-squares minimization routine (Challis, 1995). Equation A19 computes “normalized” joint angles, with the joint poses in the standing calibration trial defining the neutral “zero” angles (Davis et al., 1991). Segment angles were computed in the same fashion, with the identity matrix as \( \mathbf{R}_{\text{distal}} \).

Joint orientations in anatomical terms (e.g. flexion/extension) were calculated using a Cardan Xyz rotation sequence, which has been recommended as an international standard in biomechanics and lower limb motion analysis (Wu & Cavanagh, 1995).
REFERENCES


APPENDIX B: KINETIC MODELING DETAILS

The Visual3D model defined the joint center locations using the standing calibration trial marker positions. The hip joint center was 25% of the distance from the ipsilateral greater trochanter to the contralateral greater trochanter (Weinhandl & O’Connor, 2010). The knee joint center was the midpoint between femoral epicondyles, and the ankle joint center was the midpoint between the malleoli. Segment inertial parameters were calculated using regression equations from Dempster (1955) and the Hanavan (1964) model.

Resultant joint kinetics were calculated using a recursive inverse dynamics scheme. The body segments were “unlinked”, and a resultant force and moment were defined at each end of the free segment. The Newton-Euler equations of motion for any segment were:

\[
\sum \bar{F} = \frac{d}{dt} \bar{m} \bar{v} = \bar{m} \bar{a} \quad \text{(Eq. B1)}
\]

\[
\sum \bar{\tau} = \frac{d}{dt} \bar{I} \bar{\omega} = \bar{I} \ddot{\omega} + \bar{\omega} \times (\bar{I} \bar{\omega}) \quad \text{(Eq. B2)}
\]

where \( \bar{F} \) and \( \bar{\tau} \) are general forces and moments acting somewhere on the segment, \( \bar{v} \) and \( \bar{a} \) are the linear velocity and acceleration of the segment’s center of mass, \( \bar{\omega} \) and \( \ddot{\omega} \) are the segment’s angular velocity and acceleration, \( \bar{m} \) is the mass of the segment, and \( \bar{I} \) is the inertia tensor for moments of inertia about the segment’s center of mass.

By expanding the left-hand side of Eq. B1, the equation for the resultant force \( \bar{F}_R \) at any joint in an open chain of segments is:

\[
\bar{F}_R = \sum_{i=1}^{n} \bar{m}_i \left( \bar{a}_i - \bar{g} \right) - \bar{F}_{GRF} \quad \text{(Eq. B3)}
\]
where $m_i$ is the mass of the $i^{th}$ segment distal to the joint, $\bar{a}_i$ is the center-of-mass acceleration of that segment, $\vec{g}$ is the gravitational acceleration, and $\vec{F}_{GRF}$ is the GRF. From Eq. B2, the resultant moment $\vec{\tau}_R$ at the same joint is:

$$\vec{\tau}_R = \sum_{i=1}^{t} \left( \vec{\tau}'_i + \vec{r}_i \times m_i \left( \bar{a}_i - \vec{g} \right) \right) - \vec{\tau}_{GRF} - \left( \vec{r}_{GRF} \times \vec{F}_{GRF} \right)$$

(Eq. B4)

where $\vec{r}_i$ is the vector from the joint to the center of mass of the $i^{th}$ distal segment, $\vec{F}_{GRF}$ is the vector from the joint to the center of pressure, and $\vec{\tau}_{GRF}$ is the ground reaction moment. $\vec{\tau}'_i$ is the inertial moment of the $i^{th}$ distal segment. Equation B4 calculates $\vec{\tau}_R$ in the global (lab) reference frame. Results were then expressed in the LCS of the distal segment.

REFERENCES

