THE EFFECT OF SUSTAINED STATIC KNEELING ON KNEE JOINT GAIT PARAMETERS
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INTRODUCTION

Epidemiological studies have identified kneeling as an occupational risk factor for knee joint osteoarthritis (KOA) [1], but direct biomechanical evidence for this relationship is lacking. A possible mechanism attributed to KOA onset is that prolonged static kneeling compromises the integrity of the knee joint ligaments resulting in knee joint instability, which would manifest itself as altered ambulatory loading profiles (Fig. 1). As a preliminary exploratory investigation, the purpose of this study was to investigate the effect of sustained static kneeling on knee joint gait parameters.

Figure 1: A hypothesized pathway for the relationship between static kneeling and KOA.

METHODS

Ten healthy male subjects (24.1 years ± 3.5, 1.80 m ± 0.06, 77.66 kg ± 7.68) with no history of knee injury volunteered for this study. Each subject’s dominant leg and pelvis were instrumented with non-collinear infrared-emitting diode (IRED) clusters on rigid discs. The discs, with three IREDs each, were placed on the lateral side of the dominant foot, shank, and thigh, and over the posterior pelvis. Virtual markers from anatomical bone landmarks were also recorded via motion capture in the neutral posture using the tip of a six-IRED probe. The anatomical landmarks of interest on the dominant lower limb included the head of the first and fifth metatarsals, the medial and lateral malleoli, and the medial and lateral epicondyles, as well as bilateral heads of the greater trochanter and anterior superior iliac spines.

After a familiarization period, subjects performed ten walking trials at a self-selected normal pace over a force plate (AMTI, Watertown, MA, USA) embedded in the floor. Kinematic data was collected using two three-camera Optotrak Motion Tracking Systems (Northern Digital Inc., Ontario, Canada). A 16-channel Optotrak Data Acquisition Unit (Northern Digital Inc., Ontario, Canada) was used to collect the force plate at 1000Hz and motion data at 75Hz. Subjects then performed a static deep knee-flexion kneeling protocol consisting of three bouts of ten minutes of kneeling, each separated by a five minute seated rest period. Subsequently, a set of ten walking trials were performed after a one-minute rest and equipment verification period. The location of the anatomical virtual markers were then re-determined.

The ground reaction force and motion data were processed using Visual 3D software (Visual 3D, version 3.9, C-Motion Inc., MD, USA). The anatomical landmarks were used to define the proximal and distal ends of their respective anatomical segments, allowing for a single lower limb model to be created for each subject. A second-order lowpass Butterworth filter was used to filter the kinematic (6 Hz cutoff) and kinetic (25 Hz cutoff) data. Euler angles were used to calculate the
knee flexion angle (KFA) and knee adduction angle (KAA), and inverse dynamics was used to calculate the external knee adduction moment (KAM) and knee flexion moment (KFM) (normalized to body mass). Visual 3D’s automatic gait detection function was used to extract the data from the stance phase of the gait cycle that occurred over the force plate, and these data were time normalized to 100% of the stance phase.

For each outcome measure (KFA, KAA, KAM, and KFM) statistical analysis involved calculating the within-subject root mean squared difference (RMSD) between the waveforms for each of the trials and the mean of the trials for each condition (pre-kneeling (PRE) and post-kneeling (POST)). The average RMSD was then determined for each subject. Additionally, the RMSD was calculated between each of the POST trials and the mean of the PRE condition. A paired t-test was used to determine if there was a significant difference between the RMSD calculated within respective conditions versus the RMSD of the POST condition against the mean of the PRE condition.

RESULTS AND DISCUSSION

No statistical difference was found between the RMSD calculated within conditions for the KFA, KAM, and KFM (Fig. 2) indicating that the variances within these outcome measures for the POST condition were similar to those during PRE gait, where the latter is considered to be the normal subject-specific stance-phase profile and therefore serves as the control condition. However, a significant difference was observed in the RMSD between these two conditions for the KFA (p<0.05), with an increase in RMSD in the POST condition. This indicates an increase in KFA variance in the POST gait trials. It is important to note, however, that this comparison does not address the shape of the waveforms, only their variance within each condition.

To test for differences in amplitude, the waveforms from the POST gait were compared with the average waveform of the PRE waveforms. The RMSD increased significantly for all outcome measures (p<0.05). This indicates a significant deviation in the POST condition from the PRE condition. While this measure does not speak to the direction of the deviation, this study sought only to determine if kneeling caused a change in gait waveform patterns. Indeed, we have shown that the within subject waveform variance does not change within conditions, but that it does increase in the POST condition relative to the PRE condition, indicating that static kneeling does have at least an acute effect on gait measures.

![Figure 2: RMSD averaged between subjects as calculated both between and within conditions for a) flexion and adduction angles and b) flexion and adduction moments. ( * = p <0.05)](image)

CONCLUSIONS

As changes in gait parameters have been associated with both knee joint instability and KOA (as reviewed in [2]), we believe that the findings of this study are evidence that the proposed pathway linking static kneeling to KOA may be viable and warrants further in-depth investigation.

REFERENCES


ACKNOWLEDGEMENTS

We would like to acknowledge the support and assistance of the Human Mobility Research Centre, Queen’s University, Kingston, ON.