LOWER LIMB JOINT MOMENTS CONTRIBUTIONS TO FORWARD PROGRESSION DURING GAIT IN THE ELDERLY

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SUMMARY

The purpose of this study was to use induced acceleration analysis to quantify the contributions of lower limb joint moments to center of mass forward progression in elderly gait pattern. Three healthy and active subjects (72.67±4.04y), with no gait pathology and no history of falls in the previous year, were tested. An 8 segments model (feet, shanks, thighs, pelvis and trunk) was built and optimized through inverse kinematics. Variables computed included spatial-temporal gait variables, lower limb joint angular displacements, lower limb joint moments and induced accelerations generated by lower limb joint moments on center of mass forward acceleration. It was verified that: (1) ankle joint moments are the largest contributors to forward progression, meaning an active plantarflexor push-off; (2) the magnitude of the induced accelerations generated was somewhat lower comparing with those reported for young adults; (3) the swing limb joint moments did not contributed to forward center of mass acceleration.

INTRODUCTION

Falls have been reported as a serious problem among elderly people[1] and lower limb muscle weakness, as well as gait and balance deficit seem to assume a key role on discriminating elderly fallers[2].

Kinematic changes, like the decrease in stride length, stride velocity, relative swing duration and in ankle range of motion (ROM), have been reported for the elderly[3,4]. These kinematic changes have been associated with a decrease in ankle plantarflexor joint moment of force and power[3,4].

Also, as been suggested that the effect of age was more than a reduction in motor abilities, causing a redistribution of joint torques and powers, which emphasized the hip extensors and deemphasized the knee extensors and the ankle plantarflexors compared with young adults[5].

With these studies, authors were able to report qualitative descriptions of the strategies used to compensate for neuromuscular losses that occur with aging. However, it is not possible to quantify how much center of mass (CoM) acceleration is produced by each joint moment. On the other hand, induced acceleration analysis (IAA) allows the direct quantification of a joint moment contribution (or muscle force) on the acceleration of each body segment and has

proven to be a powerful clinical assessment tool[6,7]. This technique is based on the principles outlined by Zajac and Gordon (1989), who have proven that the joint moments produced by muscles that span a certain joint will generate acceleration in all body joints. Until now, IAA has not been used to analyze elderly gait pattern. Therefore, the purpose of this study was to use IAA to quantify the potential changes in the contributions of lower limb joint moments to CoM forward progression (forward CoM acceleration) in elderly gait pattern.

METHODS

Three healthy and active subjects, two women and one man, with more than 65y (72.67±4.04y), no neurologic or other condition that would affect their gait pattern and without any history of falls in the previous year, accept to participate in this study. Immediately prior to data collection, all participants were informed about the study, accept to participate and signed the informed consent. The Ethics Committee of Faculty of Human Kinetics approved the study protocol.

Gait kinematics and kinetics was collected with a Qualisys Track Manager system (Qualisys AB, Gothenburg, Sweden) with 12 infrared, high speed cameras (Qualisys Oqus 300, Qualisys AB, Gothenburg, Sweden) working at a frequency of 200 Hz and synchronized with two Kistler force plates (9281Be 9283U014 Kistler Instruments Ltd, Winterthur, Switzerland). Subjects were asked to walk naturally, at a self selected speed. Prior to data collection training trials were done so that the subjects would become comfortable with the task.

Two trials from each subject, in which both feet contacted the force plates (starting with the left foot), were selected to be analyzed. A fourth order Butterworth filter was used for both kinematic and kinetic data. Filter cut-off frequencies were determined by analyzing the Fast Fourier Transform of each marker position/time curve. Foot markers and force plate data filter cut-off frequency was 10Hz (the same value was applied based on the work done by Van Den Bogert et al, 1996), while a 6Hz cut-off frequency was used for the rest of the markers. Data processing was performed through a continuous pipeline developed under Visual 3D software (Professional Version v4.80.00, C-Motion, Inc, Rockville, USA). An 8 segments model (feet, shanks, thighs, pelvis and trunk) was built and optimized through inverse kinematics[10]. Computed gait
variables included spatial-temporal variables, lower limb joint angular displacements and lower limb joint moments. IAA was processed based on the method stated by Kepple et al (1997), being the forward acceleration data only evaluated when the combined ground reaction force obtained from the two force platforms was anteriorly directed (~35-55% of the gait cycle (GC)). For IAA the foot was fixed to the floor during foot flat and allowed to rotate about the centre of pressure for the rest of the time[11]. The accuracy of the model was measured for each subject by computing the absolute differences between the CoM acceleration derived from the force platform and the one induced through the model. The mean anterior-posterior error ranged from 0.06 to 0.07 m/s². For all the subjects the mean errors were less than 5% of the total range of accelerations.

RESULTS AND DISCUSSION
Study participants had a mean age of 72.67±4.04y and a Body Mass Index of 28.1±5.0 Kg/m². The results obtained for typical gait variables are in accordance with those stated in the literature[3,4] for the elderly. This means a reduction in stride length, stride velocity, Ankle ROM, horizontal reaction force peak and ankle moment of force peak at push-off, and an increase in hip ROM, when comparing with the reported values for young adults[4].

As mentioned before, it was verified that the effect of age was more than a simple reduction in motor abilities, causing also a redistribution of joint torques and powers, which emphasized the hip extensors and deemphasized the knee extensors and the ankle plantarflexors compared with young adults[5]. Our IAA results (figure 1), however, showed that despite of the typical kinematics and kinetics observed, ankle plantar flexor joint moment was the largest contributor for forward acceleration and this contribution starts before the push-off phase (~45-65% GC, thus starting on ~50% of forward acceleration interval), when the plantarflexors are acting eccentrically.

This fact was also reported by Kepple et al (1997) when testing adults. The same author also concluded that the knee joint moment may too contribute to forward acceleration during the push-off. Nevertheless, in our study, this contribution is not observed. In fact, knee joint moment produced a negative CoM acceleration, indicating that probably this joint moment has a different role, other than the generation of forward progression.

The verified contributions to progression from all the other joint moments of force of the support limb, and from gravity, were relatively small. Also, the swing limb almost did not contribute for forward progression (figure 2).

**Figure 2:** Mean (n=3) induced horizontal accelerations generated by joint moments of the swing limb on CoM during forward acceleration interval (~35-55% of total GC)

**CONCLUSIONS**
The purpose of this study was to use IAA to quantify the potential changes in the contributions of lower limb joint moments to CoM forward progression (forward CoM acceleration) in elderly gait pattern. It was verified that: (1) ankle joint moments are the largest contributors to forward progression, which is in agreement with Winter (1991) that states that forward velocity is mainly generated by an active plantarflexor push-off; (2) the magnitude of the induced accelerations generated was somewhat lower comparing with those reported for young adults[6]; (3) the swing limb joint moments did not contributed to forward CoM acceleration.

**REFERENCES**
2. Rubenstein LZ, Age Ageing. 35:S2:ii37-ii41, 2006