Intersegmental Effects of an Ankle Foot Orthosis on Joint Rotation

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Introduction
Ankle Foot Orthoses (AFO) are used to control knee and hip motion in pathological gait. AFO delivery relies on artisan skills of the provider; trial and error methods in setting the AFO joint range of motion are common. Systematic quantification of control due to brace joint constraint has not been reported. An ultimate goal of this study is to develop a simple, clinically viable method to determine optimal range of motion settings for AFOs. An intersegmental dynamics analysis (IDA) has been applied to quantify the effect of an AFO on joint motion during walking. IDA objectively relates the effect of changes in joint moments (due to the brace) on global joint rotations (Kepple et al., 1997). This can facilitate better understanding of interaction between the brace and the limb and the internal rearrangement of joint moments to compensate for the effects of the brace (Esquenazi et al., 2000). Synergy patterns can be clearly visualized. This can assist in making the bracing process more scientific.

Methods
An able-bodied test subject was fitted with an AFO. The brace joint was initially set to offer no resistance to ankle motion to assess the baseline effect of brace weight and comfort. The subject walked with this “free” brace; 3D gait data were collected. The same AFO was then fixed in five degs. of plantarflexion and the subject walked to acclimate to this condition. Testing was repeated; walking speed was maintained for both conditions. Three subjects are enrolled; two have completed the testing.

The human body was modeled as a seven segment kinematic chain. Equations of motion allowed calculation of the contribution of joint moments to global joint motion. Measured ground reaction force data were incorporated into the IDA computation rather than an artificial constraint on the foot.

Results & Discussion
Data are reported from one representative run from each condition for a single test subject walking in the locked (“test”) and free (baseline) brace conditions. See key kinematic and kinetic changes in Table 1. All comparisons are made to the baseline condition. If a variable is shown as increased, denoted by the plus (+) symbol, this implies that it was higher in the locked ankle than in the free ankle condition.

<table>
<thead>
<tr>
<th>Joint</th>
<th>Early Stance (0-20%)</th>
<th>Mid Stance (20-40%)</th>
<th>Late Stance (40-60%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Moment</td>
<td>Angle</td>
<td>Moment</td>
</tr>
<tr>
<td>R. Hip</td>
<td>+ Ext</td>
<td>± Fle</td>
<td>+ Fle</td>
</tr>
<tr>
<td>R. Knee</td>
<td>+ Ext</td>
<td>+ Ext</td>
<td>± Ext to 30%</td>
</tr>
<tr>
<td>R. Ankle</td>
<td>+ DF</td>
<td>No chg. to 15%</td>
<td>± PF to 35%</td>
</tr>
</tbody>
</table>

Table Notes: 1) All moments are presented as “internal” or that which the body had to produce.
2) All timings are presented relative to Right stride since the brace was on the R. leg.
3) Differences in moments of <2Nm or in angles of <2° are presented as “No chg.”

IDA is a method to relate the effect of a single joint moment to the motions produced throughout the body. For the current analysis, the differences in the effect of the lower extremity joint moments on rotation of the braced leg knee and hip joints are compared for two brace conditions. Early, mid, and late stance phase are analyzed. Representative data from early stance are presented in Figures 1&2. The goal was to clarify global effects of the brace on hip and knee rotation and the body response to the brace.
**Early stance.** The brace provided resistance via a dorsiflexion moment as the ground reaction force pushed the ankle into peak plantarflexion at footflat. The brace moment was larger than that normally provided by the muscles during this period [~27Nm at ~12% R. Stride vs. ~20Nm in baseline]. The effect of the prevailing dorsiflexion ankle moment at this time is to accelerate the knee into flexion (Figure 1). The increased moment due to the brace produces an additional 146 rad/s/s flexion acceleration at the knee. Additional knee flexion is not necessary and in fact undesired for this subject at this time. The body response countered this by increasing the knee extension moment [peak 53Nm at ~15% R. Stride vs. 40Nm in baseline]. The increased knee extension moment produces an additional 199 rad/s/s extension acceleration, more than offsetting the brace effect (Figure 1). The prevailing hip extension moment provides a small but significant amount of knee extension acceleration at this time. In the locked ankle condition, the hip moment changes from strong extension [20Nm] to about 8Nm flexion. Its contribution to the knee rotation acceleration similarly changes from providing extension to providing flexion. This change produces an additional 93 rad/s/s knee flexion acceleration (Figure 1). The “TotVGE” or external moment is a composite moment due to ground force, gravity, body positioning and velocity effects. Taken together, the effects of the ankle, knee, hip and “TotVGE” moments nearly perfectly balance each other out (146-199+93-49 = -9). The brace and hip produce extra knee flexion acceleration, while the knee responds with increased extension acceleration. The external moment lessens its large contribution to knee flexion resulting in an extra 49 rad/s/s knee extension acceleration. A similar chain of explanations for the effects of joint moments at the hip is now presented. 

Slightly more hip flexion acceleration [18 rad/s/s] results from the increased dorsiflexion moment (brace). The increased knee extension moment produces over three times more hip extension acceleration [58 rad/s/s] than the increased hip flexion due to the brace. Again, additional hip extension is not necessary in this case, and may adversely affect the body center of mass progression. It appears that the body responds by reducing the hip extension moment to balance the extra extension acceleration produced by the knee moment. An additional 42 rad/s/s hip flexion acceleration is seen. Together, the ankle, knee and hip moment changes produce almost the same hip rotation acceleration as in the baseline. But the balance between components is much different across conditions. Significant internal rearrangement or compensation was found at the knee also. Visualizing these effects helps to explain brace function and the synergies between the body and the brace for each brace range of motion setting.

**Midstance.** Increased plantarflexion, decreased knee extensor and increased hip flexor moments are noted. The increased plantarflexion moment accelerates the knee into extension [158 rad/s/s] while the knee and hip moderately balance this effect. The external moment produces a change of nearly 80 rad/s/s (340 in locked vs. 260 in free) and represents the largest balance to the increased knee extension acceleration due to the brace. The vertical ground reaction force only changes about 1% at this time.
making it an unlikely explanation for this difference. The A/P component of the ground reaction changes from about 6N anteriorly directed to about 21N posteriorly directed. The strong knee flexion effect of this can be visualized. Intuitively, this is also consistent with delayed progression of the center of pressure that may result from a brace locked in plantarflexion. At the hip, the increased ankle plantarflexion moment produces additional extension acceleration. However this change is almost perfectly balanced by the effects of changes in the hip and knee moments. The external moment has only a minor effect.

Late Stance. A decreased ankle plantarflexor moment is consistent with the brace motion limitation. Decreased knee extension and hip flexion moments are also noted. The change in ankle and knee moments produce less knee extension acceleration while the hip moment and external moments produce less flexion acceleration. Once again, the actual change in knee rotation acceleration is very small at this time, but the internal distribution used to achieve it appears significantly different compared to the baseline. In this case, a reduction of all major components is noted in the locked brace, indicating less opposing effects. This may be indicative or more efficiency, control, or increased synergy.

Synthesis. Two points of clarification about IDA analysis should be noted. The magnitudes of the rotation produced by the joint moments appear high because they represent the effect of each moment acting on an unconstrained body. Only the effect of segment inertias but not external constraints (ground) are included. The total effect of the ground reaction, gravity, and segment velocities are considered separately from the effect of the joint moments. These effects are combined into a single factor called the TotVGE or external moment. The sum of the effects of the bilateral hip, knee and ankle joint as well as the external moment reproduce the measured change in rotational acceleration for each joint studied. This serves as one necessary check of analysis accuracy. The IDA method used does not require a forward dynamics analysis nor does it depend on the accuracy of mathematically modeling the foot-floor interaction to compute the shown effects (Talaty et al., 2001). Error introduced by the latter is not entirely clear. The tradeoff is that the presented method sacrifices clarity in terms of the magnitude of the effects calculated.

Returning to the interpretation of early stance, it is unclear why the knee seemed to over compensate by producing more extension acceleration than the brace produced flexion. It appears that the knee moment change could have been more moderate allowing little or no contribution from the hip. Instead, the knee overcorrected for the effect of the brace forcing the hip to contribute. This may indicate a transient accommodation response. Data presented were taken after about 15-30 minutes of walking in the test brace condition. More time may be necessary to develop a more “efficient” response pattern. It is also possible that the effect of the knee moment is significant elsewhere in the performance profile, such as the contralateral hip or the forwards propulsion of the body. These effects can be studied as well with the given methods. Based on IDA data, a synergy or performance index that assesses how much the effects of the joint moments oppose each other or conversely work together is currently being developed.

Knowledge of internal balance of the effect of joint moments may assist in selection of the most suitable brace for a patient. IDA explicitly calculates the motion of joints throughout the body due to a single joint moment. This was used to clarify the effect of an AFO on hip and knee joint motion. An AFO not allowing dorsiflexion is often prescribed to prevent stance phase knee buckling in patients with weak knee extensors. Care must be taken in applying a generic rule. For example, in the case presented the brace was shown to produce significant additional knee flexion acceleration in early stance; this was stabilized by the knee extensors and further mediated by hip control.

References

Acknowledgements. This work was supported in part by the Albert Einstein Society. Our sincere thanks go to Oliver Woods, Jr. and Barbara Hirai for their continuing help in this project.