Feed-forward torque control for plantar flexion support


Abstract—Recent studies suggest that a reduction of the plantar-flexion muscle EMG can improve the efficiency of human locomotion. In this study, an attempt is made to reduce the maximal EMG of the gastrocnemius medialis and soleus by providing a feed-forward torque support around the ankle. The strategy only requires the ankle kinematics to define the appropriate support torque. The maximal amount of support applied was equal to 30% of the total ankle torque of a human during steady state walking. Five untrained subjects walked on a treadmill at 4km/h and completed 3 test sessions of about 45 minutes each. The final EMG measurements showed a maximal reduction of 30% for the soleus and a slight increase of the gastrocnemius medialis EMG pattern for the case where 50% of the maximal support was applied. The tests showed a significant change in the ankle angle patterns when walking with maximal support. The findings suggest that feed-forward torque control is capable to reduce the soleus muscle EMG and could possibly increase the walking efficiency of human locomotion.

I. INTRODUCTION

Of the lower limb joints, the ankle creates the highest torques during locomotion [1]. From the physiological point of view, the plantar flexors are the main contributors of this torque and are essential for the velocity and support of the center of mass. One way to increase the efficiency of human walking is by supporting those muscles with exoskeletons [2], [3], [4].

Over the last decade, several ankle exoskeletons have been developed and studies were executed which allow to gather a basic knowledge of muscle-tendon behavior under the influence of exoskeletons.

An EMG controlled ankle exoskeleton developed by Sawicki and Ferris had a positive effect on the net metabolic costs during walking at different speeds [5], [6]. The maximal plantar flexor assistance achieved 34–40% (test subject dependent) of the total lower limb mechanical work and was able to decrease the net metabolic costs by 7–26% respectively.

In another study done on the same exoskeleton, Kao et al. [7] suggest that the human under the influence of an exoskeleton prioritizes the conservation of the torque pattern over the kinematics pattern during steady state walking. In their studies, subjects walking with powered assistance had a 36% reduction of the soleus muscle activity and maintained their ankle moment pattern while the ankle kinematics pattern changed.

An EMG controlled exoskeleton from Kinnaird and Ferris achieved a reduction of the gastrocnemius medialis by 12% while the soleus muscle activity was reduced by 27%. Their findings suggest that EMG controlled exoskeletons present a productive tool for promoting motor adaptations in humans [8].

Sawicki and Collins [9] presented a concept study for an optimized mobility assistance during walking. One of their key design objectives was the capability to provide a torque pattern which is similar to the normal ankle joint moment.

Kong presented a fictitious gain controller which includes the brain in the control loop [10]. The exoskeleton uses ground reaction force inputs to determine the human torque around the ankle (i.e. the inverse dynamics). With this knowledge, a low pass feed-forward controller determined the amount of support given. Experimental results using this strategy where only shown for the lower arm muscles and not for the plantar-flexion muscles.

In the attempt to reduce the muscle activity of the gastrocnemius medialis, Lintzen presented an approach using a combination of an oscillator-based phase detection and a spring-damper predictive controller [11], [12]. The controller parameters where optimized in simulation on beforehand and only require the ankle encoder data as input signal. The results showed that the proposed support strategy was unable to reach its goal and led to the conclusion that the subject was still in a learning state at the end of the test. The data analysis also showed a deviation between the predicted and measured ankle angle, resulting in a non-optimal set of parameters.

The analysis of free walking subjects with a motion caption
system revealed overall similar ankle angle patterns but significant differences in magnitude per individual. Further, the angle-torque relationship is nonlinear during one complete gait cycle.

Recapitulating all those findings, the following problems were identified:

- Most powered ankle exoskeletons require ground reaction forces or EMG as input, making the devices complicated and susceptible to noise. EMG sensors are sensitive to the location they are attached to and require normalization.
- The ankle angle varies in magnitude for each individual and could potentially change under the influence of active support. Furthermore, the angle-torque relationship of the ankle is nonlinear. These factors make it difficult to rely purely on the ankle angle to implement a working support strategy.

The goal of this research is the definition of a support strategy that reduces the muscle activity of the soleus and the gastrocnemius medialis by a significant amount. Eventual changes in the ankle kinematics and other muscle EMGs involved in human locomotion are also monitored.

The support torque chosen is equal to a fraction of the average normal ankle torque of the human during steady state walking and is independent of EMG inputs. Instead, the strategy relies on a phase detection to define the appropriate amount of feed-forward support. The hypothesis is that this approach will allow the wearer to reduce its soleus and gastrocnemius medialis muscle activity.

All experiments in this study are executed with the PAFO ankle orthoses (Figure 1): a one degree of freedom exoskeleton developed by Lintzen [12].

For the verification of the strategy, several fully abled test subjects walked on a treadmill at constant speed. The effectiveness of the provided support is evaluated with the normalized EMG of the leg muscles.

II. METHOD

A. Subjects

Ten healthy male subjects (body mass $77 \pm 8$ kg; $26 \pm 3$ years of age) volunteered to participate in this study. The subject did not have any walking impairments nor had a lower limb surgery over the past year. All tests where approved by the local committee of ethics at TU Delft.

B. Support Torque

The maximal supplied support torque is expected to be at 30% of the actual human ankle torque, which is approximately 0.6 Nm/kg. The shape of the torque remains unchanged over the phase and only varies in magnitude based on the weight of the person wearing the exoskeleton. The implementation of this feed-forward method is achieved by a look up table and only requires the phase as input signal. Figure 2 shows the normalized average human ankle torque (Nm/kg) and power (W/kg). Both show a global maximum at 50 and 55% respectively.

C. Phase Detection

The detection of the phase transforms the wavelike ankle angle signal into a linear increasing sawtooth signal (between 0 and 100).

The kinematic pattern of the ankle during steady state walking is consistent and repetitive; so, even though the magnitude of the angle and velocity change per individual (not allowing the determination of a fixed global maximum or minimum), the local maxima and minima remain at approximately the same gait cycle percentage. Figure 3 shows the average ankle kinematic patterns of a complete gait cycle.

The adaptive oscillator presented by Ronsse et al. [13] uses the property that any cyclic trajectory $\theta(t)$ can be estimated by a sum of sinusoids with different amplitudes $A_i$, angular frequencies $\omega_i$ and offsets $\phi_i$.

$$\theta(t) = \sum_{i=1}^{k} A_i \sin(\omega_i t + \phi_i)$$

The method creates a number of oscillators that are summed up to reproduce and the estimated trajectory $\dot{\theta}$:

$$\dot{\theta} = \sum_{i=1}^{k} A_i \sin\left(\frac{2\pi \times \phi_i}{100}\right)$$

Latter tries to match the initial trajectory $\theta(t)$ as much as possible and has an additional condition: $d\phi_i < d\phi_{i+1}$ The gait phase $\phi_{base}$ corresponds to $\phi_1$. 

Figure 2. Average human walking torque and power pattern during walking of the right and the left side ankle.
The starting point of the defined phase does not match with the beginning of the stance phase and has to be shifted. The shifting term is determined with consistent properties of the human gait kinematics. (Figure 3, 4). In the following section, the process to achieve this approach is explained step by step.

To start off, the initial phase \( \phi_{\text{base}} \) is divided in the middle, such that each half contains a local velocity maximum. In the exceptional case, the global velocity maximum is at the 0 or 50 percent mark or within a 5% perimeter; the phase is shifted by 10%.

In a second step, the acceleration patterns of the left and the right ankle are used to identify the half containing the swing phase. In the case, the local left side acceleration minimum priors the right side’s, the respective half also contains the beginning of the swing phase.

Figure 3. Average ankle angle patterns. The blue line represents the side where the phase is to be detected. The red line represents the global minimum angle at the 50% gait phase mark. The black line shows the maximal velocity closest to the beginning of the swing phase at the 62% gait phase mark.

Figure 4. Phase detection process: 1. Define the initial phase \( \phi_{\text{base}} \) with the adaptive oscillator and the ankle angle as input. 2. Filter and differentiate the angular velocity and acceleration of the ankle. 3. Split the phase in two halves (red dashed line) and verify that the overall maximal velocity lies outside the gray dashed area. If this is not the case, shift the current phase by 10%. 4. Determine the minimal acceleration positions in the current gait phase \( \phi_{\text{base}} \) of both sides (left \( \phi_L \) and right \( \phi_R \)) and in each half. Subtract the left minimum from the right one \( \Delta_{R-L} = \phi_R - \phi_L \). 5. Find the maximal velocity position \( \phi_{\text{mv}} \) in the half where \( \Delta_{R-L} \) is positive. The beginning of the stance phase occurs 62% before \( \phi_{\text{mv}} \).
The lower level control consists of a PI controller with an adaptable feed-forward gain and is optimized for a low frequency environment (from 0-15 Hertz). The controller uses the previous gait cycles to determine a feed-forward gain which allows the handling of the nonlinear properties of the air muscle.

E. Protocol

Each test subject performs three tests on separate days (minimum of 45 hours break in between each test session) to reduce the influence of learning effects. In each test, the subjects walked at a speed of 4.0 km/h. One subject absolved the first training session at 4.5 km/h. The Pafo exoskeleton was adjusted in height and positioning to allow an optimal functioning for each test wearer.

During each test day, the subjects walked with 0, 15 and 30% of torque support for 15 minutes each. In the 0% support mode, the exoskeleton controller tries to minimize the recorded ankle torque. In the active support mode, 15 respectively 30% of the average ankle torque where feed-forwarded to the human.

We recorded the ankle angle, actual (measured) torque support given, feed-forwarded torque support and the corresponding phase during each test session. On the final day, the subjects additionally walked without exoskeleton for 5 minutes to record free walking EMG behavior. The EMG muscle activation of the soleus, gastrocnemius medialis, tibialis anterior, vastus lateralis, biceps femoris and the semitendinosus were recorded. The EMG sensor locations where determined with the Seniam protocol (www.seniam.org).

F. Data Analysis

We used the data of the final minute of each test phase and for each session. With the help of the phase detection, the average step cycle profiles with corresponding variances where deducted for the ankle kinematics (angle, velocity), the ankle torque and the EMG data. All the data was collected at 1000 hertz.

The EMG signals where bandpass filtered (4th order) in between 30 to 400 Hz, followed by an IIR notch filter at 50 and 100 Hz to eliminate unwanted noise from electrical power supply. In a final step, the signal was rectified and low-pass filtered (2nd order) at 6 Hz. To allow a comparison in between test subjects, the maximal EMG amplitude of the 0% support case (baseline) was normalized to 1.

To calculate the angular velocity and acceleration, the encoder signal is first low pass filtered (2nd order) at 10 Hz and then differentiated respectively.
G. Statistics

The significance levels of the results were evaluated with a one sided t-tests. Latter compared the average maxima and minima over all the test subjects and for each case with each other.

The standard deviation calculations determined the learning effects over the three test sessions.

III. RESULTS

A. Learning effect

Over the course of the three test sessions, the subjects adapted to walking with active support. All subjects reached the anticipated maximum support torque of 0.59 Nm/kg in the second and the last session. The overall maximal standard deviation decreased from ±0.1354 Nm/kg in the first session over ±0.0929 Nm/kg for the second session to ±0.0854 Nm/kg for the final session. These findings indicate an adaptation and learning effect to walking with the exoskeleton.

B. Kinematics

Looking at the final session in detail, the average of the minimal ankle angle rose from -0.05 rad at 0% over 0.0662 rad at 15% to 0.1036 rad at 30% support (Figure 8).

The average cycle time decreased minimally with increasing amount of support (Figure 7). The variations between subjects where however too big and the average changes too small to reject the null hypothesis of the t-tests.

The difference between the maximal and minimal angle over a gait cycle was calculated for each subject. The mean over all the subjects of latter value showed a significant decrease in between the 0% support case and the 15 and 30% support case (Figure 9). In numbers, the average angle difference decreased from 0.43±0.06 rad to 0.334±0.036 rad (23% reduction) at 15% support and to 0.29±0.042 rad/s (32% reduction) at 30% support. The standard deviations decreased for the active support cases compared to the 0% case.

The average maximal velocity decreased significantly in between the support cases with the exception of the first and second 0% support case (Figure 10). In numbers, the average maximal velocity decreased from 3.695±0.955 rad/s to 2.2982±0.28989 rad/s (37% reduction) at 15% support and to 1.727±0.15 rad/s (53% reduction) at 30% support. The standard deviation decreased with increasing amount of support.

C. Electromyography

The maximal amplitude of each case was than meaned over all the subjects with the corresponding standard deviation (Figure 13).

Soleus: The analysis of the maximal average soleus EMG amplitude showed a significant reduction between the 0% and the 30% support case (P=0.009). The EMG decreased from 1±0 to 0.6975±0.14 which corresponds to a 31% reduction.

Gastrocnemius Medialis: The analysis of the maximal average gastrocnemius medialis EMG amplitude showed a significant change between the 15% and all the other support cases. The EMG between the 0% and the 15% case increased by 9.7% from 1±0 to 1.097±0.068 (P=0.0177). The EMG between the 15% and the 30% case changed by 14.7% from 1.097±0.068 to 0.95±0.09 (P=0.0234). The EMG between the 15% and the second 0% case changed by 22.7% from 1.097±0.068 to 0.878±0.133 (P=0.0173). The EMG between the 15% and the free walking case changed by 24.7% from 1.097±0.068 to 0.84±0.207 (P=0.0493).

Tibialis Anterior: The maximal average tibialis anterior EMG amplitude did not change significantly.
Figure 11. Maximal mean amplitude EMG with corresponding standard deviations. The significant P value cases are listed above.

Figure 12. Average kinematic and torque pattern of the last minute of each support case with corresponding standard deviations.
**Vastus Lateralis:** The maximal average vastus lateralis EMG amplitude did not change significantly.

**Biceps Femoris:** The maximal average biceps femoris EMG amplitude did not change significantly.

**Semitendinosus:** The analysis of the maximal average gastrocnemius medialis EMG amplitude showed a significant change between the 0% and the 15, 30 and second 0% support cases. The EMG between the 0% and the 15% case decreased by 30% from 1 ± 0.708 ± 0.224 (P=0.0473). The EMG between the 0% and the 30% case decreased by 37.5% from 1.5 ± 0.6253 ± 0.277 (P=0.0415). The EMG between the 0% and the second 0% case decreased by 35.2% from 1 ± 0.64 ± 0.24 (P=0.0327).

---

**IV. SINGLE SUBJECT ANALYSIS**

To get a better understanding on the changes that occur with different support cases and sessions, the average kinematic, torque and EMG patterns with corresponding standard deviations of a single subject are presented.

**A. Kinematics and torques**

The analysis of the ankle angle, angular velocity and torque data indicates that the standard deviations decreased over the course of the three sessions (Figure 13).

On the third session, a clear increase of minimal ankle angle (50% gait phase) when walking with active support compared to the 0% support case was detected. The first and second 0%
Figure 7. Average cycle time reduction between the different support cases for all the support cases with corresponding standard deviations.

Figure 8. Average ankle angle kinematics for different support cases. The control 0% case was increased compared to the first one, which seems logical regarding the corresponding maximum angle results.

B. Electromyography

The analysis of the EMG data indicates that only the soleus and tibialis anterior change by a considerable amount under different support influences (Figure 14).

The maximal soleus EMG amplitude (45% gait phase) is the highest for the free walking case and the lowest for the 0% support case. The second 0% case was not significantly different from the first one.

Figure 9. Average between the minimal and maximal angle amplitude (max. angle - min. angle over the gait cycle) for every support case with corresponding standard deviations. The outcome showed a significant decrease at the 15 and 30% support case compared to the 0% support case. The standard deviations decreased as well. The horizontal lines represent the P value results of the t-test comparisons for the significant cases. The t-tests resulted in a significant difference between the feed-forward torque supported cases and the 0% support case. The second 0% case was not significantly different from the first one.

Figure 10. Average maximal velocity for every support case with corresponding standard deviations. The maximal average velocity decreased significantly at the 15 and 30% support case compared to the 0% support case. The standard deviations decreased with increasing amount of support. The horizontal lines represent the P value results of the t-test comparisons for the significant cases. The t-tests resulted in a significant difference between all the states. Only the second 0% case was not significantly different from the first one.
increasing amount of support. The standard deviation becomes phase is lowest at the free walking case and increases with an increasing amount of support. The standard deviation becomes also larger as the amount of support increases.

V. DISCUSSION

The results of this study support the hypothesis that it is possible to lower the plantar flexion muscle activity with a feed-forward torque controller. Compared to 0% support case, the subjects tested showed a soleus EMG reduction of 31% for the 30% support case. The gastrocnemius medialis EMG increased significantly for the 15% support case compared to the other cases. This is an unexpected result, and could possibly be related to an ongoing learning effect. Another hint to this conclusion is the fact that the second 0% support case EMG was lower than the first one and the 30% case. The differences in EMG amplitude where much less prominent for the gastrocnemius medialis than for the soleus. A possible explanation for latter result lies in the fact that, unlike the soleus which is exclusively responsible for the ankle movement, the gastrocnemius is also involved in the knee movement.

The tibialis anterior EMG increased at the beginning of the stance phase (0-40% of the gait cycle) when walking under active support (Figure 13). I conclude therefore, that the provided torque support is not parallel to the human ankle torque in that region. The human therefore counteracts to the exoskeleton and actively tries to maintain his torque pattern with his tibialis anterior muscles (as a sort of controller).

The desired feed-forward torques where met and had little variance.

The kinematic pattern however changed considerably. The difference between the minimal and maximal ankle angle changed by up to 32% (Figure 9) while the maximal velocity decreased by 53% at maximal support compared to the 0% case (Figure 10). These results conclude that the subjects reduced the movement of their ankle and walked more on their toes during active feed-forward support. The findings suggest that the human prioritizes changing his kinematic pattern over his torque pattern.

The reduction of the maximal velocity came along with a reduction of variance. This result suggests that the proposed strategy provides a sort of guidance while walking. As a consequence, an application in the rehabilitation field might be possible.

The average soleus EMG at the second 0% support at the end of the test was lower than the first 0% support case. A possible explanation for this phenomenon is that the steady state walking was not yet reached for every test subject after 3 sessions. With additional sessions, this difference could possibly decrease.

Recalling the reduction of the soleus EMG in combination with the absence of a significant increase of all the other muscle activities at maximal support, the proposed support strategy has the potential to increase the walking efficiency during steady state walking.

VI. FUTURE RESEARCH

Due to the dynamic properties of the current setup, the maximal torque support was restricted to 30%. To explore the full potential of the feed-forward support strategy, a higher maximal torque support is required.

The results have shown an elevation of the tibialis anterior EMG in early stance phase due to high torque support. This unwanted effect could possibly be prevented by lowering the torque support in that region, such as proposed by Bregman et al [14].

In the case of the intention to lower the gastrocnemius medialis muscle activity, the current exoskeleton setup should be extended with an active knee torque support.

REFERENCES