Muscle Function During Human Gait Near The Preferred Transition Speed

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If we started walking normally and gradually increased our speed, we will eventually get to a certain point where we will prefer to run than walk. The speed at which this occurs is called the preferred transition speed. Much work has been done on the reasons for the walk to run transition, yet still very little is known about why we do it.

Previously, Thorstensson and Roberthson [1], while unable to pinpoint a distinct reason, suggest that subjective feelings of relative comfort based on previous experience and peripheral receptor information and central locomotion controlling neural activity cause humans to change gait. Hreljac [2], while eliminating kinetic factors, suggested that maximum angular velocity and maximum angular acceleration in ankle dorsiflexion during walking were critical factors in the transition to running [2,3]. Furthermore, Hreljac suggests that muscular stress, or fatigue, in the dorsiflexor muscles appears to precipitate the gait transition and that by transitioning to a run, these localised stresses can be removed and the work load shared with the upper leg muscles, despite being more energetically costly [4]. In a later study, Neptune and Sasaki computed muscle forces using a nine degree-of-freedom model and dynamic optimization techniques [5]. They found that by transitioning to a run at higher speeds, the ability of the ankle plantarflexors to generate force is greatly improved due to the fact that the muscle was then operating on a better portion of its force-velocity curve.

The hypothesis being tested is that the preferred gait mode at a given speed is chosen to minimise overall muscle stress. A non-physiological static optimisation approach utilizing inverse dynamics was used in this study to solve for muscle forces given experimental kinematic and ground force reaction data.

The skeletal model, based on earlier work by Anderson [6], describes the mechanical structure of the lower half of the human body (Figure 1). It contains ten segments and 23 degrees of freedom. The pelvis contains three rotational and three translational degrees of freedom. The femur and pelvis is joined by a 3 degree of freedom ball joint, the tibia and femur by a 1 degree of freedom hinge joint, the hind foot and tibia by a 2 degree of freedom universal joint and the toes and hind foot by a 1 degree of freedom hinge joint. The head, arms and torso are modeled as a single segment and are joined to the pelvis by a 3 degree of freedom ball joint. Fifty-four muscle actuators are included in the model (Figure 2). The muscle paths are symmetric, with 24 muscles actuating each leg, and 6 muscles actuating the back. Each actuator is modeled as a three element muscle in series with a linear tendon [7]. The mechanical behaviour of the muscle is described by a Hill-type contractile element (CE) which models the muscle's active force-length-velocity properties, a series-elastic passive element (SEE) which models the muscle's active stiffness, and a parallel-elastic element (PEE) which models the muscle's passive stiffness [7].

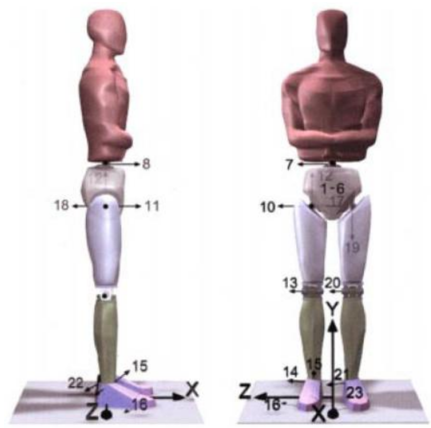

Figure 1: Sagittal and frontal plane views of the model skeleton. From Anderson & Pandy [8].

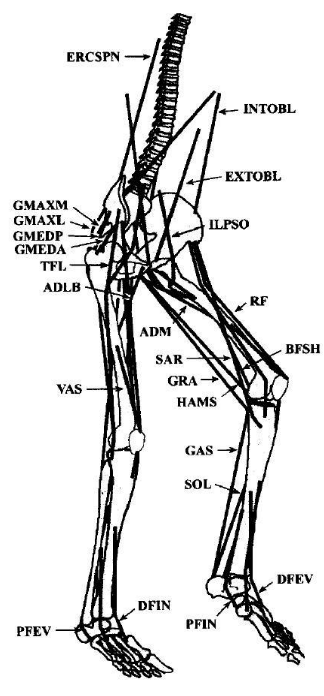

Figure 2: Schematic showing some of the muscles in the model. A total of 54 musculotendinous actuators controlled the model. From Anderson & Pandy [8].

Four subjects were chosen for gait analysis on the basis of their proximity to the model's parameters (model weight: 71kg, model height: 180cm, subject mean weight: 67.2 kg, mean height: 181 cm). Out of the four subjects tested, one subject's data was chosen for subsequent preferred transition speed analysis on the basis of closeness of kinematics to the model for normal walking presented by Anderson and Pandy [8]. The preferred transition speed of the subject was determined by allowing the subject to initially walk on a motorized treadmill at a speed lower than the preferred transition speed (about 1.5m/s). The speed was then incrementally increased by the operator while keeping the display hidden from the subject. At each speed, the subject was asked to walk for 30s, and then to run for 30s, and asked which was the more comfortable gait. The speed was then increased until the subject indicated that they were indifferent between running and walking.

Reflective markers were placed on the subjects to measure the kinematics during gait using high speed cameras and motion capture software (Vicon Systems, Oxford, UK). Kinematic data was captured at 100 Hz. By knowing the three dimensional locations of the markers on each of the ten segments that make up the skeletal model, the joint angles in each spatial direction can be uniquely determined [6]. Ground reaction forces and their point of application were measured using two ground mounted force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA). Kinetic data was captured at 1000 Hz.

An inverse dynamics approach was used to calculate the torques at each joint required to produce the measured kinematics, which then was used to estimate the forces in each of the muscles that span the joints.

However, in the human body, there are more muscles spanning any given joint than the number of degrees of freedom allowed at that joint. This excludes the possibility of a unique solution for the set of muscle forces, and thus the problem is indeterminate. To overcome this redundancy problem, optimisation techniques were used [9]. A linear relationship between the muscle's maximum force and the corresponding activation signal was assumed in a method to evaluate muscle force in a non-physiological setup.

The increase in ankle joint torques was found to be more significant as compared to the more proximal joints such as the hip joints as gait speeds increased (Figure 3). This is most likely because the increase in ground reaction force that is attributed to increased gait speed is directly applied to the foot. Consequently, this is one of the primary reasons why the ankle plantarflexor force is believed to play a primary role in gait.

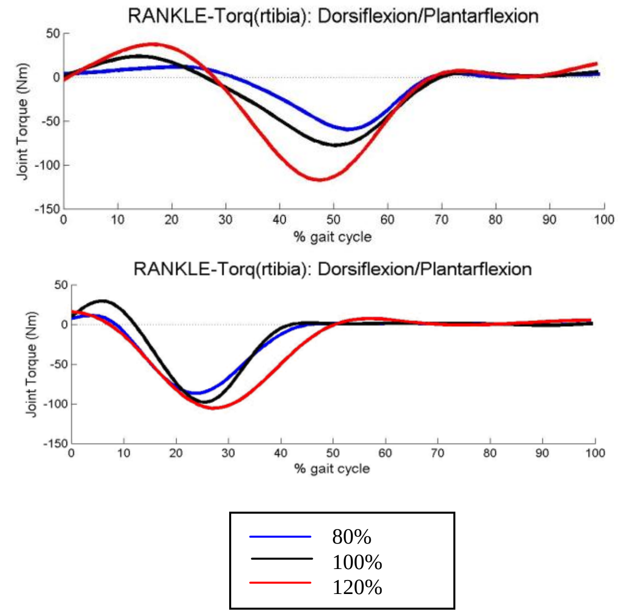

Figure 3: Ankle flexion joint torques in the right foot during: (top) Walking, (bottom) Running

The main difference between walking and running is that the ankle dorsiflexor muscle forces are reduced by a factor of two, and are not significantly utilized in running. This concurs with the findings by Hreljac, which pointed to the muscular stresses in these groups of muscles at speeds approaching preferred transition speed leading to running as the preferred choice of gait. It is believed that due to the flight phase, there is a reduced need for dorsiflexion to achieve toe clearance [2].

Additionally, once the gait mode is switched to running, the activation of the ankle plantarflexor, Soleus, is shifted earlier in the gait cycle to coincide with the earlier propulsion phase, while the magnitude of the force production in Soleus increases by approximately one third. Force production also increases with increased running speed. However, the ankle plantarflexors (Soleus and Gastrocnemius muscles) produce a peak muscle force of around 900N and 600N respectively, regardless of running speed. It can therefore be concluded that Soleus is the main contributor to the increased vertical ground reaction force in running, further clarifying the findings of Neptune and Sasaki [5].

At the preferred transition speed, it is believed that muscle lengths and shortening velocities are such that muscle physiology is necessary for a proper understanding of the gait transition [5]; however, the non-physiological static optimisation analysis presented does show that the muscles involved in ankle plantarflexion are critically involved (Figure 4).

This study could form the basis for a physiology-based static optimisation approach, or alternatively, can be used in comparison for newer methods, such as Neuromuscular

Tracking (NMT) where closed loop control systems techniques enable the mathematical model to track desired kinematics and joint torque trajectories with unwanted disturbances (such as non rigid skin and muscle movement) [10]. This future research may benefit studies into muscle function by yielding more accurate force estimations in less time than conventional methods.

The ultimate benefit of research in this field is that by having more efficient procedures to determine accurate muscle forces and joint loads during human locomotion, increasingly complex and realistic models can be used to more precisely pinpoint affected areas in patients with muscle disorders in a clinical setting, and help identify possible treatment on a case to case basis. The advances in computational speed further aid the cause to improve the quality of life for these patients.

Note: The simulations and plots were all produced in Matlab, which forms the computing basis of this study. The flexibility, power, and usability of Matlab make it the software of choice in many engineering disciplines.

Acknowledgements: The author would like to thank James Balfour and Wei-Lee Tjio for their help with gait data collection and processing.

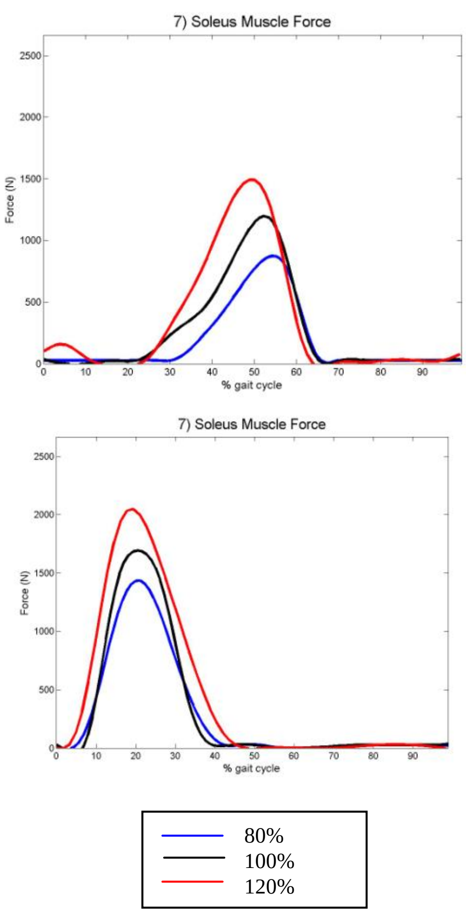

Figure 4: Simulated muscle forces in (top) Soleus during walking, (bottom) Soleus during running at speeds of 80%, 100% and 120% of the preferred transition speed.

1) A. Thorstensson, H. Roberthson, *Acta Physiol Scand.* ***131**, 211* (1987).
2) A. Hreljac, *Gait & Posture,* **1**, 217 (1993)
3) A. Hreljac, *Journal of Biomechanics.* **28** 669(1995).
4) A. Hreljac, *Human Movement Science.* **14** 205 (1995)
5) R. R. Neptune, K. Sasaki, *Journal of Experimental Biology.* **208** 799 (1995)
6) F. C. Anderson, *Thesis,* The University of Texas at Austin (1973)
7) F. E. Zajac, "Muscle and Tendon: Properties, Models, Scaling, and Application to Biomechanics and Motor Control," *CRC Critical Reviews in Biomedical Engineering*, J. R. Bourne, ed., (CRC Press, Baca Raton, 1989), Vol. 19, pp. 359–411
8) F. C. Anderson, M. G. Pandy, *Journal of Biomechanical Engineering.* **123** 381 (2001)
9) F. C. Anderson, M. G. Pandy, *Journal of Biomechanics.* **34** 153 (2001)
10) A. Seth, M. G. Pandy, *Journal of Biomechanical Engineering.* **40** 356 (2006)