Design and Analysis of a Harness for Torso Force Application in Locomotion Interfaces

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Abstract. This paper presents a new design of a harness for torso force application in locomotion interfaces. This mechanical coupling between a user's body and the force feedback device is a key aspect for the effective exploitation of this class of devices. Issues of compliance, backlash, force distribution, and comfort have been considered in the new design. Experimental results show that subjects adjust their hip torque depending on the exact point of force application along the back.

1 Introduction

One of the newest classes of mechanical interfaces for virtual reality systems are locomotion interfaces, which seek to permit natural locomotion though virtual environments. The ability to explore virtual environments by foot is hypothesized to increase the send of immersion and realism, and even to improve the appreciation of spatial layout and distance [15].

The two main types of locomotion interfaces to date are based on treadmills or on independent foot platforms [10]. At the moment it appears that treadmill style devices permit the most natural forms of locomotion, and are closer to being commodity devices that could proliferate. Alternative approaches to treadmill designs include generalized tilting platforms [13], deformable belts to represent stairs [14], and two-dimensional belt surfaces [3,11].

Besides adding complexity to the treadmill platform, another approach has been to add torso force feedback to achieve a variety of locomotion effects. The main example is the Sarcos Treadport (Figure 1), which is comprised of a 6-by-10 foot belt on a tilting platform, a CAVE-like visual display, and a mechanical tether attached to the user's back via a whole-body harness [8]. The mechanical tether has six sensed degrees of freedom to measure user position and orientation, which are employed to control the treadmill belt speed, turning, and motion through a virtual environment. In addition, pushing and pulling forces can be exerted along the linear axis roughly in the forward direction. The effects achieved to date include:

Fig. 1. The Sarcos Treadport

1. Collision display, modeled as viscoelastic forces exerted on the user similar to haptic interface contact with a surface.
2. Inertial force display, in which the artificial inertial forces are exerted on the user based upon measurements of belt acceleration [2]. This makes treadmill running more similar to ground running, because users do not actually accelerate their bodies on a treadmill.

3. Frontal slope display, by exerting an artificial gravity in a direction parallel to the treadmill belt [7]. Although this duplicates slope presentation through ordinary treadmill tilting mechanisms, there are advantages to not moving the treadmill platform, including speed of response and no interference with the visual display.

In addition, qualification studies were performed that showed side force application could display side slopes on a different system and using passive force application [9]. Although not in the context of virtual reality systems, there have been other examples of torso force application on treadmills.

1. Passive vertical support forces have been employed to simulate reduced gravity environments [5].
2. Vertical forces in the downward direction using elastic bands were proposed for treadmill exercising in the zero-gravity conditions of space flight [12].
3. Horizontal forces have been applied using passive elastic bands to increase horizontal ground reaction forces and raise the metabolic cost of running in reduced gravity environments [1].

2 Harness Models

In all of these examples, there has to be some harness worn by the user to which the external force source is applied. When the forces are constant as in the passive application systems, the harness design can be simpler because the harness backlash need be taken up only at the beginning of force application. For time-varying force applications such as the simulation of varying slopes or changes in inertial force, any backlash or excess compliance will result in reduced force bandwidth, fidelity, and stability. Another potential complicating factor is the need for multi-dimensional force application: frontal, side and vertical forces.

In the present study, harness design is considered from a standpoint of frontal force application only. A particular goal is to reduce the backlash in the design resulting from straps and soft tissue contact, and to improve the mechanical coupling to the torso, while addressing important issues of human factors.

2.1 Backpack-style harness

The Treadport system originally employed a backpack style harness to connect the external mechanical device to the subject's body (Figure 2). The tether is connected to a metal plate, embedded in the harness back side. This solution allows high flexibility of adaptation over different users by simply adjusting the strap lengths to fit different sizes. However, several drawbacks are present.

Fig. 2. Backpack-style harness with mechanical connection to the active tether.

The stiffness of the mechanical coupling with the tether strongly depends on the way the different straps are positioned and tightened. Experimental measurements, obtained by positioning several markers on the user's body and the tether and applying different forces, showed a stiffness ranging between 4.6 and 13N/mm. Due to errors introduced by surface skin mobility and the motions of the subject, these preliminary values are only indicative. More precise measurements, obtained for instance by using a rigid mannequin, would still be unsatisfactory, as the absence of the natural body compliance would remove a key factor in the behavior of the overall mechanical coupling.

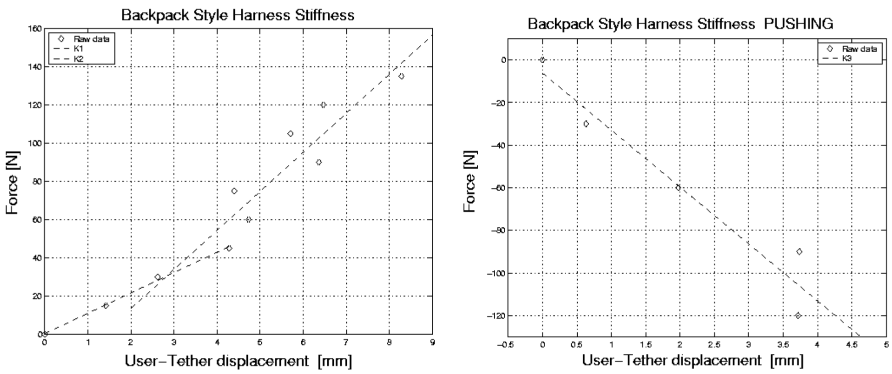

Fig. 3. *Left:* Bilinear interpolation for the estimate of a backpack-style harness stiffness in pulling action. *Right:* behavior in pushing.

The behavior of the connection provided by this kind of harness can be identified: for low force values, the straps are not tightened by the external force and the stiffness is lower. As the force increases the straps get progressively tighter, partial sliding between them occurs, as the inter-strap friction is overcome by a threshold applied force, and the user body is partially "squeezed". At higher forces the inter-strap sliding occurences decrease and the body squeezing action reaches a balance. As a consequence the felt stiffness increases.

This behavior is shown in the left side of Figure 3, where the force-displacement profile is approximated through bilinear interpolation. Figure 3 is only indicative of the general harness behavior because the slope of the fittings ($K_1 \approx 10$[N/mm], $K_2 \approx 20$[N/mm]) and the slope changing force ($\approx$ 30N) strongly depend on such factors as strap positioning and tightening, and the percentage of soft mass in the user's body. Another drawback is that the harness behaves in two clearly different ways during pushing and pulling.

- During pulling, the force is applied through the straps that roll around the user's hip, crotch, shoulders, chest and stomach. The back plate departs from the user on the basis of the straps tightening and user's torso compliance (in particular at the abdomen), producing a backlash. This is due to the back straps that partially align themselves in the tether direction, when a pulling force is applied, thus producing a perceptible displacement.
- During pushing, all the force is applied directly on the back of the user, through the contact and pressure of the back plate. In this way the straps' tightness and compliance is basically unimportant and the coupling stiffness only depends on the

softness of the user's back, which shows a lower change from subject to subject. As a consequence in the pushing action (negative force) the harness has a stiffer and more constant behavior as shown in the right side of Figure 3, where the force slope is (K3 ≈ 26.5 [N/mm]). In this case the straps' initial tension is unimportant, creating only a small initial gap, if the straps are very poorly tightened, that is immediately canceled when the pushing action starts.

From the user's perspective, the difference in behavior means feeling a simpler, cleaner, force on a restricted area on the back of the hip, when the tether is pushing, and a complex body-squeezing (chest and stomach areas) and pulling (shoulders and hip front) action when it is pulling. The difference between these two conditions is enhanced by the thermal feeling of fresh air flow when the harness departs from the user's back during pulling.

2.2 Mechanism-based harness

In order to investigate how the interface-user mechanical coupling affects the sensation of motion in the Treaport locomotion interface, a metal-based prototype harness has been created (Figure 4). The basic principle in the creation of this new harness has been the elimination, from the back side, of every fabric-based part that perceptibly increases the coupling compliance in the pulling phase. The whole back side of the new harness is composed of aluminum tubing and plates. A spherical joint at the back of the hip and two revolute pairs at the level of the upper body center of gravity allow the necessary DOFs for free movement of the torso during walking and running. This passive mechanism is again fixed to the user's body by several straps that in this case are applied only to the front side of the body, thus avoiding the compliant effect mentioned earlier during pulling.

Fig. 4. A mechanism-based harness.

To increase the comfort and best reference and fit to the structure of a subject, some formable plastic sheets may be used, even if they do not actually enhance consistently the mechanical performance of the harness. The assembly of the metal components comprising the hip, back and shoulders links can be easily rearranged to fit different users in a good range of heights and sizes.

In order to reduce the compliance due to the deformation and squeezing of the user's body itself, the restraints have been rolled around bony parts: hip, crotch, and chest. A first solution for the upper body, comprising two straps directly rolled around the shoulders was rejected because it decreased blood flowing at the arms on the long term. Indeed two main arteria and veins were pressed by the shoulders straps. The present solution uses two straps that are crossed on the user's chest. In Figure 5 the final locations of the restraints are reported with respect to the human body skeleton and vascular system.

Since the back part of this new harness consists of aluminum parts only, its stiffness is much greater than in previous solutions. However some compliance due to unavoidable body deformation still exist and is difficult to determine by experimental measurement since the errors introduced by subject motion and skin deformation are of the same magnitude of the tether-body relative displacements that are subject to measurement. Nevertheless practical use of the new harness shows a better coupling with the user and coherence in the pushing and pulling actions. Indeed, since the back structure is stiff and isostatically connected to the body, the forces that are applied at the hip and upper body level are always a constant fraction of the external tether force.

3 Harness Mechanics vs. User Biomechanics

For predicting the force to be applied to a user for inertial force display or slope display, the user has simply been modeled as a single mass [2,7,9]. With the backpack-style harness, no other assumption is possible because the distribution of forces over different straps is unknown. With a stiff and isostatic structure, instead, the fraction of the external force that is applied to the hip and upper body is constant and easily determined by the static equilibrium of the tether, hip and upper body forces. This passive sharing is driven by the system geometry only, and thus it is completely and easily defined. Figure 5 identifies how a harness can distribute forces to multiple body points: the external force F is shared between upper and lower body (F_u, F_h); the upper force is further shared between lower torso and left/right shoulders (F_{u1} , F_{u2} , F_{u3}), by static balance of the harness upper plate as governed by the geometry (a,b,c,d) .

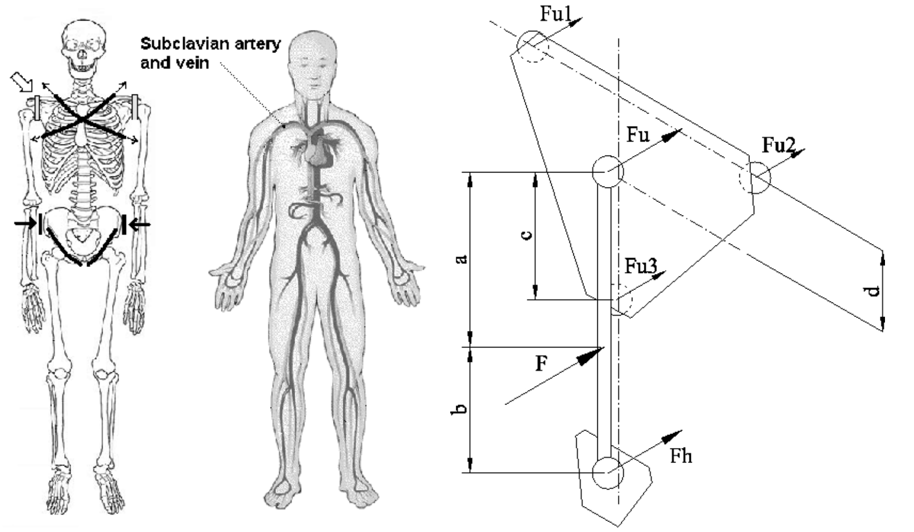

Fig. 5. *Left:* Positioning of the harness restraints on the human body with respect to skeletal structure and blood vessels. Final straps positioning (black), rejected shoulders restraints (white). *Right:* passive sharing of an external force F to multiple points.

Since the new harness allows to vary the point of application of the external force in the vertical direction, by sliding the attachment point of the tether along the back link of the harness (Figure 4), it has been used to vary the application point of the external force along the torso. This means applying an increasing fraction of the overall force to the upper body, reducing the fraction at the hip level.

Several tests were performed over 4 different subjects: in each one the subject was walking at a constant speed of 1m/s while the force was varied in the range ±120 N with 20N increments. After applying the different forces at a given application point, the point was varied. The experiment was repeated for four positions at increasing distances from the hip: pos. 1 (lower) to pos. 4 (upper). The subjects' posture was monitored by a Northern Digital Optotrack 2030 3D tracking system.

A first analysis showed that good linearity exists between applied force and upper body sagittal angle in every single trial, particularly in the case of pulling (positive force). Standard deviations from linearity, for upper body angles of -10 to 30

[deg], were in the range of 0.47 to 1.75 [deg] for pushing (mean 0.88 [deg]) and of 0.42 to 2.0 [deg] for pulling (mean 0.89 [deg]).

Figure 6 represents, for subject 1, a plot of the applied force versus upper body angle, in the sagittal plane, with respect to a vertical reference. Single trial data points are represented together with the respective linear interpolation. It is clear that by increasing the elevation of the application point with respect to the hip, the same force produces, in average, a larger upper body angle. The trend of the successive trials is also stressed by arrows in the figure. A similar average increase of the upper body angle with the elevation of the application point showed for subjects 2 to 4 (Figure 8).

However a more precise relation between the force application point and upper body posture assumed by the walking subject can be found by considering the torque produced at the hip by the external force. In Figure 7 the upper body sagittal angle over the different trials is reported with respect to the applied hip torque. In this case the linear interpolations for the different trials match very closely, independent of the position of the application point. This suggests that, in this kind of force application, the upper body posture control is based on hip torque, not just on the magnitude of the exerted force.

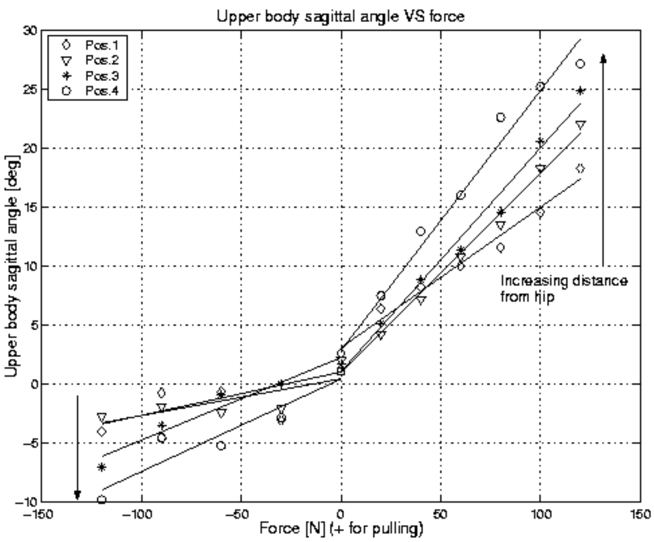

Fig. 6. Effect of changing the application point of the external force: upper body sgittal angle versus applied force for increasing distances from the hip (Pos1 to Pos4).

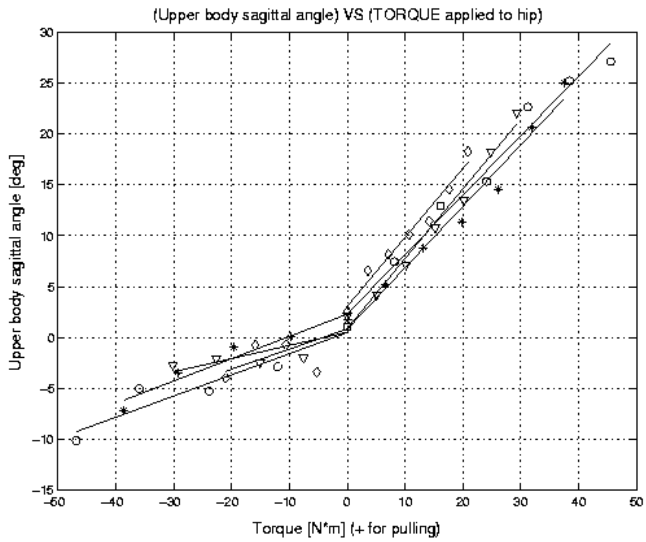

Fig. 7. Effect of changing the application point of the external force: upper body sagittal angle versus moment produced at the hip by the exerted force.

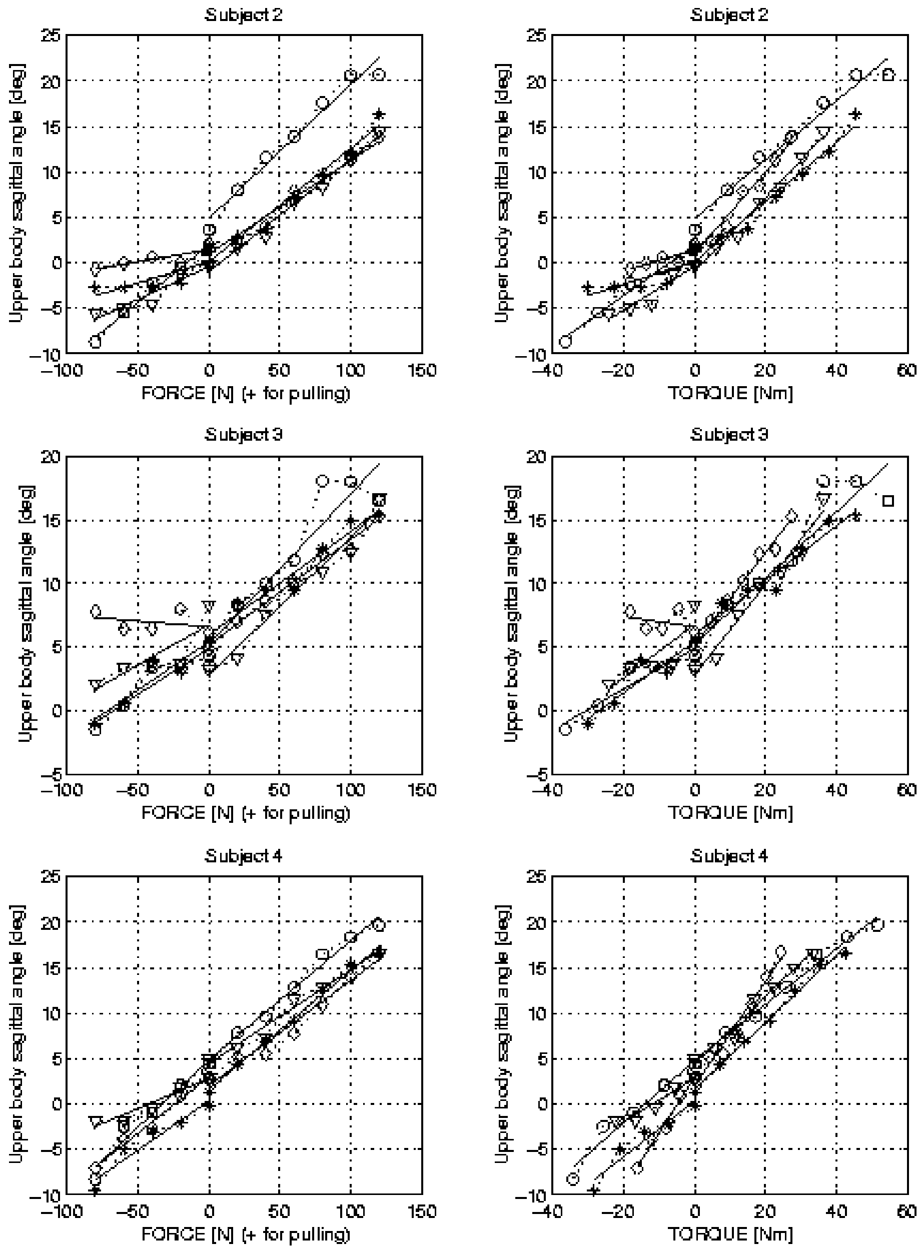

Fig. 8. Plots for subjects 2 to 4 of the upper body angle with respect to applied force or hip torque. Data legend is as in Figure 6.

Table 1. Means of plot values of upper body sagittal angle [deg] vs. Force and hip torque, over all individual subjects and trials.

F_{pull}	Subj.1	Subj2	Subj.3	Subj.4	F_{push}	Subj.1	Subj.2	Subj.3	Subj.4
Pos.1	10.23	7.14	10.12	8.95	Pos.1	-1.05	0.38	6.94	-1.98
Pos.2	11.15	6.47	9.27	10.63	Pos.2	-1.68	-3.30	4.22	0.16
Pos.3	12.43	7.59	10.81	9.43	Pos.3	-1.3	-1.75	2.37	-4.00
Pos.4	16.06	13.70	12.30	12.72	Pos.4	-3.08	-3.10	1.98	-1.10
T_{pull}	Subj.1	Subj2	Subj.3	Subj.4	T_{push}	Subj.1	Subj2	Subj.3	Subj.4
Pos.1	10.23	7.14	10.12	8.95	Pos.1	-1.05	0.38	6.94	-1.98
Pos.2	8.38	4.61	7.66	8.94	Pos.2	-1.43	-2.57	4.86	0.88
Pos,3	7.55	4.61	8.87	6.06	Pos,3	-1.16	-1.00	3.59	-2.15
Pos.4	9.07	9.29	8.75	8.56	Pos.4	-1.67	-0.57	3.43	2.01

Table 1 provides a numerical comparison over all subjects of the mean upper body angle vs. force or torque. Each column represents the mean angle for every subject in the successive trials. There is generally a positive trend in case of pulling force and a negative one in case of pushing force. When torque is considered for interpolation, since the resultant torque values span increasing ranges over the different trials, for comparison consistency the considered torque range has been reduced to that of the 1st trial. In this case the mean upper body angle does not show particular trends over the different trials.

4 Discussion

The new mechanism-based harness has been successful in reducing backlash in the mechanical coupling of external force to user's torso, and allows a more predictable force application to different parts of the body. This differentiated force application could be important to allow more detailed modeling of the biomechanical effects of external force on the body. Previous research has shown that a model of a human as a single mass overpredicts the amount of force that should be applied to a user. Biomechanical and psychophysical studies show that 65% of the predicted force should be appied for slope display [7], while psychological studies show that 80% of the predicted force should be applied for inertial force display [2]. A more sophisticated segmental model may account for these reduced force amounts. Particularly in the display of virtual slope, we may reduce the necessary simulation force by adopting an application point that is closer to the upper body center of mass.

While the upper body control strategy based on hip torque mentioned above holds for most of the subjects, it is affected by experience/confidence in using the locomotion interface. The average standard deviations over all subjects of the slopes of the angle vs. torque plots are $S_{k1t} = [0.61_{push}\ 0.19_{pull}]$, considering a mean value of 1 over each data set. There is a consistent behavior in case of pulling force, but less so for pushing force particularly at the lower force application point. This confirms that as the hip torque variability decreases (force closer to the hip), the subjects lose sensitivity to this parameter, and the usual upper body control strategy fails. When the first trials are neglected, the standard deviations become $S_{k1t} = [0.36\;0.15]$, showing a better behavior, especially for pushing.

The importance of walking confidence on the locomotion interface has been exemplified by two more subjects, completely inexperienced, that have been asked to perform directly the experiment. Their behavior showed initial incoherence and consistent differences from the expected posture: in this phase, indeed, they were still keeping confidence with the system. These differences progressively disappeared toward the end of the trials. These subjects were both about 1.9m tall and the harness length didn't suit perfectly to them, however, despite comfort issues, the time to acquire the necessary confidence is likely much higher in presence of an active mechanical tether, with respect to walking on a simple treadmill. This should be considered if such devices are intended for applications accessed by a large number of users for very short periods of time (e.g. entertainment applications in theme parks).

For future development of the Sarcos Treadport, there will need to be added the capability to apply side and vertical forces to the user's torso. This could require more detailed models for the user and more sophisticated and definite rules for the sharing of forces over the user's limbs. For the vertical direction only, a possible harness is proposed in [12] for treadmill training in microgravity conditions on the International Space Station. The vertical force applied to an astronaut is shared between shoulders and hip by means of pulleys. It is clear, however, that if the external force can assume an arbitrary direction, these solutions cannot be easily implemented without consistently increasing the harness mechanical complexity, thus losing the benefits of this approach. The passive force sharing solution of our proposed design might be generalizable to apply defined fractions of an external single force to different parts of the body, without having to rely on more complex components.

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