Pneumo-tronic Perturbator for the Study of Human Postural Responses

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Abstract. This paper describes the design and operational principles of a device that imparts a well-controlled mechanical force or impulse, a so-called perturbation, to a pre-selected point on the surface of the human body. This perturbator will be integrated within a system aimed at measuring and evaluating human postural reaction in a clinically meaningful way. The ease of use and versatility of the device renders it suitable for manual operation but it can also be integrated in a robotized system. The hardware, control law and characterization of the perturbator are presented. Preliminary results indicate that the device is able to generate repeatable perturbations with characteristics appropriate to the intended application. Further improvements are discussed and proposed.

Keywords: postural control analysis, pneumatic actuation, human response, perturbation, human balance.

1 Introduction

Maintenance of body balance, using the so-called postural reactions, is a very fundamental neuro-motor skill necessary for the performance of everyday activities. Compromise to the postural control system may have significant consequences including an increased risk of falling, which might lead to subsequent injuries, and reduced quality of life. Therefore, a quantitative analysis of postural stability is important for assessing the various associated impairments as well as for evaluating the efficacy of interventions such as rehabilitation or pharmacotherapy [1].

Postural control involves the interaction of three sensory systems: the somatosensory system, the vestibular system [2, 3] and gaze control [4, 5]. Postural deficits may be better understood by examining the postural control response to an external perturbation (e.g. a mechanical, visual or auditory). A possible scheme of the human balance control system subject to external perturbations [6] is shown in Fig. 1. The *global sensory system* measures the current postural status (human response) whereas the *controller* is designed to reduce the error between desired and perceived posture, the latter being estimated by the *sensory integration system* that processes the information provided by the sensors. Among the several kinds of external perturbations, we focused on the mechanical ones. In these cases, the perturbation often consists of a movement of the support surface [7, 8], but it can also be applied directly to upper body segments, e.g. by pulling [9-12] or pushing [13] the trunk, shoulder or pelvis. Past approaches proposed to use an external perturbation applying bulky and rather expensive devices which limit the mobility of these systems and the ecological validity and feasibility of the clinical trials. In these applications, the point of application of the perturbation is often predictable, thus affecting the response of the patients [14]. In most of the cases, the devices described in previous studies did not allow control and/or adjustment of the force as well as the impulse (time-integral of the contact force) magnitude that were applied to the subject, making the analysis and comparison of the results difficult.

Furthermore, standardization of the measurement of postural reactions was never achieved, reflecting among other things, the diversity of technological approaches to the problem. Some solutions were based on the use of force platforms to track the center of pressure (COP) of the ground-foot interface [15, 16] whereas others applied accelerometers and inertial units [17-20], surface electromyography [21] or motion analysis [22].

This paper describes the design and function of a new mechanical perturbating device capable of imparting a targeted, force-controlled impulse to the human body. This perturbator may be carried by the examiner/clinician or secured to a stable base of support. The former option is enabled by the specific features of the device namely it is simple, lightweight, transportable and adaptable to the patient's characteristics and the clinical task at hand. The design is based on a commercial pneumatic cylinder controlled by means of flow proportional control valves. Pneumatic actuators are usually inexpensive and are suitable for the use in clean environments. Although there are several nonlinearities involved in pneumatic actuation systems, which encumber their control, a number of previously reported control algorithms [23-26] enable a high level of performance.

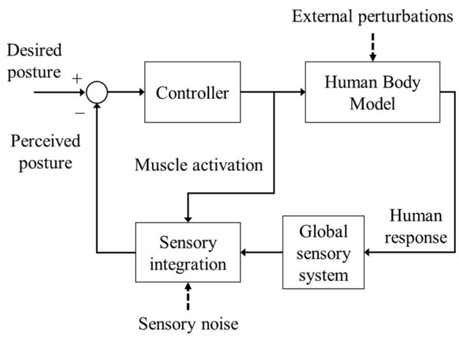

Fig. 1. Scheme of the human balance control system subject to external perturbations.

2 Materials and Methods

This study focuses on the design of an automatic perturbator (AP) used for studying dynamic balance in patients. The 'business end' of the AP which is momentarily interfaced with the human body, termed the striker, is intended to generate a controlled perturbation to the body. The main objective in both healthy subjects and patients is to analyze comparatively the response pattern to the perturbation and in the latter, to identify relevant correlates to the severity of the disorder and its variations due to intervention. In this study, the contact force between the striker and the subject body is considered to be the only controllable external perturbation while the other kinds of stimuli such as visual or auditory, are considered noise and should be limited to avoid undesired effects on the body response evoked by the perturbation.

Among the several types of mechanical perturbations proposed in the literature, very limited attention has been given to the application of a push force on the subject. Despite the simple conception of such a stimulus, the perturbation given by a push is not simple to define and to control. The perturbation generated by the AP is tailored individually to the subject in terms of timing, amplitude and direction of the stimulus. The entity of the perturbation can be defined through several physical quantities: in addition to the waveform of the contact force between the striker and the stricken body, the peak force as well as the impulse may be considered. Unlike the traditional approaches (moving base platform or cable-based systems), the AP is not limited in terms of the direction as well as the point of application of the perturbation. Two separate working configurations of the AP have been considered and analyzed:

- fixed: in which the AP is connected to a fixed-frame (e.g. a pole, or the end-effector of a manipulator) thus providing a rigid connection to the environment;
- hand-held: in which the AP may be handled by the clinician, who acts as a viscoelastic constraint towards the environment.

It should be emphasized that the second configuration represents undoubtedly an easier-to-use and compact solution compared to its fixed-frame counterpart, albeit introducing additional uncertainties to the characterization of the device, which could affect negatively the performance of the control system and the accuracy of the analysis. The proposed device and its architecture are shown in Fig. 2, constituted by the following elements:

1. a linear double acting pneumatic actuator (MetalWork, mini-cylinder Series “ISO 6432”, bore diameter 25 mm, stroke 120 mm);
2. two flow proportional control valves (CKD, 3AF2, 0-10V of analog input, working pressure range 0-0.97 MPa) and dedicated control drivers (CKD APC-23);
3. a uniaxial load cell (Dacell UMM, rated capacity about 500 N, nonlinearity, hysteresis and repeatability 0.1% of the rated capacity) and signal conditioner (DEWETRON, DEWE-RACK-4);
4. an end striker, integral with the load cell, covered by a thick layer (10 mm) of a synthetic foam material working as a deformable buffer;
5. two handles (in aluminum) used to maneuver the device;
6. a trigger button to enable the perturbation.

The complete perturbation system includes an air compressor and pressure regulator (gauge supply pressure set to 3.5 bar); a real-time system (dSPACE), Multi-Channel A/D Board DS2002 (16 bit, 32 channels) for signal acquisition and D/A board DS2101 (12 bit, 5 channels) for control generation; a Personal Computer with MATLAB®-Simulink® Real-Time (The MathWorks) and ControlDesk® (dSPACE) software.

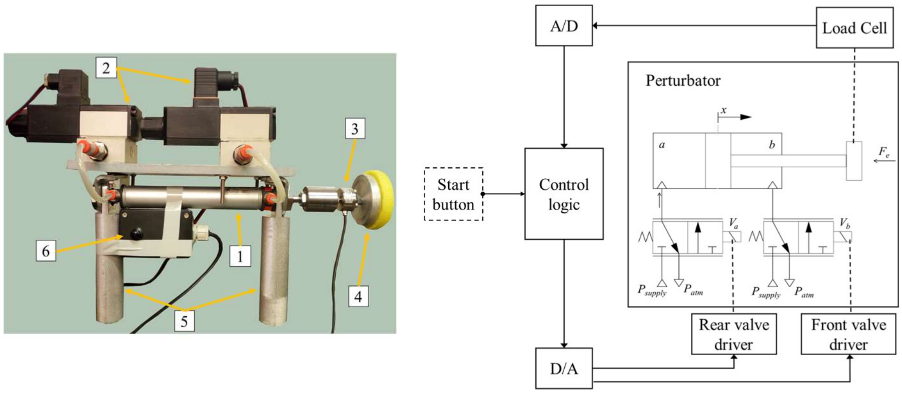

Fig. 2. Photo (left) and a scheme (right) of the perturbation system. The components highlighted are: 1 – actuator, 2 – valves, 3 – load cell, 4 – buffer, 5 – handles, 6 – trigger button.

The stroke of the pneumatic cylinder has been considered appropriate for the application, since the clinician is expected to stand near the subject, placing the end of the perturbator at short distance from the body (about 100 mm). This actuator was chosen for its large availability and low cost and it was tested to evaluate its performance in a demanding application as the one proposed in this paper. However, more expensive solutions with significantly higher dynamic performance are available (e.g. low friction metal seal cylinders), which may be tested in the future. With reference to Fig. 2, the two three-way valves have been connected respectively to the rear (a) and front (b) chambers of the cylinder, in order to control the input-output mass flow rate from both chambers. In this work, the system has been designed in order to perform the tracking of a reference contact force profile with short duration (typically in the range 250-500 ms) and amplitude in the range 20 to 100 N. Given the limited size of the actuator, these high-flow rate valves were considered appropriate for the purpose and able to fulfill the dynamic requirements of the stimulus. However, the presence of significant friction in the pneumatic actuator chosen could limit the performance and accuracy of the force tracking, mainly due to the stick-slip effect.

The valves have been placed on a base integral with the actuator's body, as shown in Fig. 2, in order to shorten as much as possible the pipes towards the chambers, thus optimizing the dynamics of the pneumatic circuit. A metal cylindrical support has been fixed at the end of the piston rod and used to carry the uniaxial load cell. The range of the load cell, considerably higher than the amplitude of the contact force, has been chosen to avoid unsafe conditions for the transducer in case of unexpected instabilities in the control logic and for characterization purposes. The sensor has been preliminary calibrated and zeroed through the conditioning software before each set of trials. The presence of a uniaxial force transducer rather than a multi-axis sensor significantly simplifies the acquisition and processing of the impact data. However, it requires appropriate care and training of the operator to avoid bending of the load cell rod during the application of the perturbation. The free end of the rod is rigidly connected to an aluminum cylindrical component (about 50 mm of diameter) covered by a deformable synthetic foam, the latter acting as the soft interface between the actuator and the body of the subject. This component distributes the contact pressure on a wider area than the sectional area of the load cell rod in order to prevent discomfort.

The acquisition of the sensor output is performed by means of a data acquisition and real-time system by dSPACE. Such system has been also used to develop and test the control logic. The sampling rate was set to 200 Hz. The implementation of the control algorithm has been enabled using MATLAB®-Simulink® through the Real-Time™ Toolbox, while the user interface for the parameter's selection and data viewer has been developed in ControlDesk®.

The working condition of the AP consists of three phases whose sequence is initiated by an external trigger (activated by the clinician). The three consecutive stages are the approach phase, the strike phase and the return phase (see Fig. 3). In the absence of an external command, the system remains in an idle condition, which represents the fourth stage of operation. For these reasons, the control logic has been conceived as a finite-state machine and implemented directly through the Stateflow® environment in Simulink®. The approach phase is triggered directly by the occurrence of the start command and consists of the extension of the piston. When the contact force exceeds a defined threshold (selected as the 10% of the reference force magnitude), the system moves to the strike stage, tracking the desired force profile. The transition to the return phase is triggered by the fall of the reference force signal, then the idle condition is reached after a time-out period of 4 s since the start command.

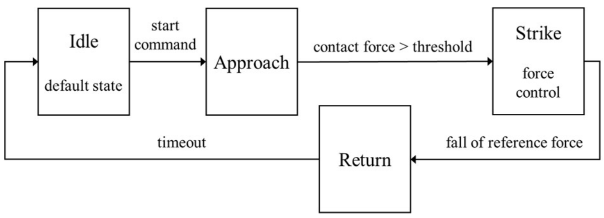

Fig. 3. General scheme of the control logic sequence.

During the approach and return phases, the system has been set to drive the control valves with constant voltage signals (open-loop) due to the inability of the controller to continuously monitor the piston position or velocity due to lack of specific transducers. Such values were selected by preliminary trials, verifying that the striker would not impact the subject's body (or a fixed target) with high speed, in order to avoid the risk of a significant overshoot in the contact force produced by the passive response of the stricken body. The resulting moving speed of the piston rod was considered sufficient for the application, always producing extension and retraction phases lasting less than 1.5 s with good repeatability. Since any overshoot in the contact force may harm the subject, it is possible to drive the piston at a very high speed only during the return stage. To perform a better control of the impact, it may be necessary to monitor the actual velocity of the rod with a transducer, thus bringing to a more complex control law. The strike phase, in which the interaction between the striker and the target has to be kept under control, has been designed by implementation of a PI controller, evaluating the error between the required force profile (in this case represented by a rectangular function defined by a duration and an amplitude) and the actual contact force measured by the load cell and filtered by a Butterworth low pass filter (4^{th} order, cut frequency 10 Hz), designed in MATLAB® (Signal Processing Toolbox™) and necessary to remove the effect of the powerline noise. The control system had to ensure the correct synchronization between the reference signal and the measured one. The tuning of the PI controller was performed by a semi-empirical iterative approach (i.e. Ziegler-Nichols) for different durations and amplitudes of the stimulation. This controller has been chosen for its simple formulation and implementation, thus not taking directly into account the nonlinearities within the system, that may require a more advanced design (e.g. sliding mode control).

The device has been characterized by performing several trials in laboratory: the dynamic performance of the system has been assessed with the actuator fixed on the work bench (configuration 1), testing contact force profiles lasting 250 and 500 ms and considering different initial distances between the striker and a fixed target (55 and 80 mm). Then, the AP was tested in a hand-held condition (configuration 2) by a skilled operator. The operator applied the perturbations while trying to keep the same initial distance between the striker and the fixed target.

3 Results of the experimentation and Discussion

Figure 4 shows the results of the first characterization trials performed with the AP in fixed (left) or hand-held (right) configuration. In both cases the tracking of the force profile seems to be insufficiently accurate, demonstrating slow transients in the initial and final phases of the strike. The overall duration of the stimulation reaches about 350-400 ms, while the amplitude of the contact force remains closer to the reference level. In both cases, however, the system produced quite repeatable results, showing a better performance in the fixed condition. This result was expected since the performance of the control system was affected by the mechanical impedance of the operator in the hand-held configuration as well as by the repeatability of targeting the specific point on the subject's body, an inherent attribute of the operator. The size of the actuator could also affect the dynamic performance of the device, thus a more compact cylinder may be necessary to get a fast-responsive system. To assess the repeatability of the trials, the time integral of each perturbation was calculated and compared with the impulse of the reference signal. For the 50 N set-level trials (reference impulse equal to 12.50 Ns, Fig. 4, bottom), in the fixed-frame and hand-held configurations, the impulse was 13.88 $\pm$ 0.05 and 12.49 $\pm$ 0.24 Ns, respectively. This outcome indicates good repeatability among the several trials. The force tracking shows acceptable dynamics, considering the step variation of the reference signal. This result was reconfirmed for the trials performed with a reference signal of 75 N amplitude (Fig. 4, bottom), as well as for different durations of the perturbation (Fig. 5).

Since in the hand-held set-up the operator cannot easily keep a constant initial distance between himself and the body target, we tested the performance of the system for different values of the initial distance (80 and 55 mm) between the (fixed) actuator and the target. The results of this analysis are shown in Fig. 6: the tracking of the force profile seems to be slightly affected by the initial distance, showing an overshoot of the contact force at 55 mm but not at 80 mm.

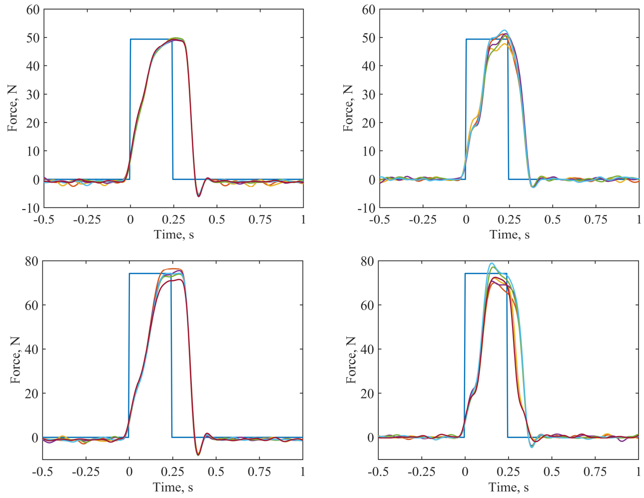

Fig. 4. Force tracking in fixed (left, top and bottom) or hand-held (right, top and bottom) configuration. The figures on top refer to a force reference of 50 N, the bottom ones to 75 N. In all trials, the duration of the stimuli was 250 ms.

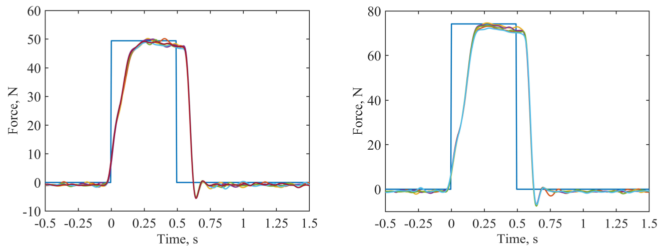

Fig. 5. Force tracking in fixed condition for 50 N (left) and 75 N (right) perturbations lasting 500 ms.

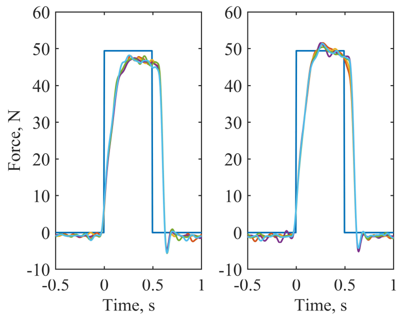

Fig. 6. Force tracking in fixed condition for different values of the initial distance between the perturbator and a fixed target: 80 mm (left) and 55 mm (right).

Figure 6 refers to 500 ms perturbations, almost coincident results were produced by 250 ms perturbations. This outcome may suggest an improvement in the design of the controller, e.g. by performing a closed-loop control of the speed of the piston rod during the approach phase, leading to a more robust control law. Preliminary tests of the AP, using a small group of healthy volunteers, has already indicated that the perturbations imparted by this apparatus are well tolerated and resulting in no feeling of discomfort.

4 Conclusion

Past approaches to imparting perturbations to human subjects during clinical postural control tests involved non-scalable systems, had a very limited number of points used for applying the perturbation, or did not allow a real-time control of the stimulus. This work presents a novel pneumatic perturbation system capable of imparting customizable force perturbations directly to the body of a subject.

The system has been conceived to control the intensity of the perturbation not only in terms of its peak force but also in those relating to more significant parameters such as the value of the force impulse. Some experimentations with the system fixed on a structure or hand-held by an operator are presented. A comparison of the results highlights the robustness of the device, given that only slight differences are noticeable by comparison of the fixed and hand-held configurations.

Future steps will involve a more robust control law, e.g. based on the monitoring of the actuator stroke and velocity, and the development of a model of the human stance control system aimed at interpretation of the body response under external perturbations. Different architectures, e.g. based on more compact pneumatic cylinders with low friction, as well as on electromechanical actuators, shall be tested to improve the tracking of the force profile.

References

1. Visser, J. E., et al: The clinical utility of posturography. Clinical Neurophysiology 119, 2424-2436 (2008).
2. Horak, F. B., Macpherson, J. M.: Postural orientation and equilibrium. In: Handbook of physiology, 253-292 (1996).
3. Horak, F. B., Nashner, L. M., Diener, H. C.: Postural strategies associated with somatosensory and vestibular loss. Experimental brain research 82, 167-177 (1990).
4. Pankanin, E., Dobosiewicz, A. M., Mietkowska, P.: Static posturography as an instrument to assess the balance among athletes. Journal of Education, Health and Sport 8(5), 216-225 (2018).
5. Gandelman-Marton, R., Arlazoroff, A., Dvir, Z.: Ocular dominance and balance performance in healthy adults. Gait & Posture 31, 394-396 (2010).
6. van der Kooij, H., et al: An adaptive model of sensory integration in a dynamic environment applied to human stance control. Biological Cybernetics 84, 103-115 (2001).
7. Farahpour, N., et al: Electromyographic responses of erector spinae and lower limb's muscles to dynamic postural perturbations in patients with adolescent idiopathic scoliosis. Journal of Electromyography and kinesiology 24(5), 645-651 (2014).

8. Grassi, L., et al: Quantification of postural stability in minimally disabled multiple sclerosis patients by means of dynamic posturography: an observational study. Journal of Neuro-Engineering and Rehabilitation 14(1):4 (2017).
9. Colebatch, J. G., Govender, S., Dennis, D. L.: Postural responses to anterior and posterior perturbation applied to the upper trunk of standing human subjects. Experimental Brain Research 234, 367-376 (2016).
10. Martinelli, A. R., et al: Light touch modulates balance recovery following perturbation: from fast response to stance restabilization. Experimental Brain Research 233(5), 1399-1408 (2015).
11. Mansfield, A., et al: Effect of a Perturbation-Based Balance Training Program on Compensatory Stepping and Grasping Reactions in Older Adults: A Randomized Controlled Trial. Physical Therapy 90(4), 476-491 (2010).
12. Mansfield A., Maki, B. E.: Are age-related impairments in change in support balance reactions dependent on the method of balance perturbation? Journal of Biomechanics 42, 1023-1031 (2009).
13. Chen, B., Lee, Y. J., Aruin, A. S.: Role of point of application of perturbation in control of vertical posture. Experimental Brain Research 235(11), 3449-3457 (2017).
14. Jiping, H., Chun-Xiang T.: Adaptive postural control for a repeated perturbation. 18th Int. Annual Conf. IEEE Trans. Automatic Control 36, 322-332 (1996).
15. Prieto, T., et al: Measures of postural steadiness: differences between healthy young and elderly adults. IEEE transactions on biomedical engineering 43(9), 956-966 (1996).
16. Ruhe, A., Fejer, R., Walker, B.: The test-retest reliability of centre of pressure measures in bipedal static task conditions - A systematic review of the literature. Gait & Posture 32(4), 436-445 (2010).
17. Moe-Nelssen, R., Helbostad, J. L.: Trunk accelerometry as a measure of balance control during quiet standing. Gait & Posture 16, 60-68 (2002).
18. Alberts, J. L., et al: Using accelerometer and gyroscopic measures to quantify postural stability, Journal of Athletic Training 50(6), 578-588 (2015).
19. Hubble, R. P., Naughton, G. A., Silburn, P. A., Cole, M. H.: Wearable sensor use for assessing standing balance and walking stability in people with Parkinson's Diseases: A systematic review. PLoS One 10(4):e0123705 (2015).
20. Heebner, N. R., et al: Reliability and validity of an accelerometry based measure of static and dynamic postural stability in healthy and active individuals. Gait & Posture 41, 535-539 (2015).
21. Henry, S. M., Fung, J., Horak, F. B.: EMG Responses to maintain stance during multidirectional surface translations. Journal of Neurophysiology 80, 1939-1950 (1998).
22. Rossi, S., et al: Compensation to whole body active rotation perturbation. Gait & Posture 39, 621-624 (2014).
23. Faudzi, A., Mustafa, N., Osman, K.: Force Control for a Pneumatic Cylinder Using Generalized Predictive Controller Approach. Mathematical Problems in Engineering 2014, Article ID 261829 (2014).
24. Driver, T., Shen, X.: Pressure Estimation-Based Robust Force Control of Pneumatic Actuators. International Journal of Fluid Power 14, 37-45 (2013).
25. Chiaverini, S., Sciavicco, L.: The parallel approach to force/position control of robotic manipulators. IEEE Transactions on Robotics 9(4), 361-373 (1993).
26. Stanisic, R. Z., Fernandez, A. V.: Simultaneous velocity, impact and force control. Robotica 27(7), 1039-1048 (2009).