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A Spherical Parallel Three Degrees-of-Freedom Robot for Ankle-Foot Neuro-Rehabilitation

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Abstract— **The ankle represents a fairly complex bone structure, resulting in kinematics that hinders a flawless robot-assisted recovery of foot motility in impaired subjects. The paper proposes a novel device for ankle-foot neuro-rehabilitation based on a mechatronic redesign of the remarkable Agile Eye spherical robot on the basis of clinical requisites. The kinematic design allows the positioning of the ankle articular center close to the machine rotation center with valuable benefits in term of therapy functions. The prototype, named *PKAnkle*, Parallel Kinematic machine for Ankle rehabilitation, provides a 6-axes load cell for the measure of subject interaction forces/torques, and it integrates a commercial EMG-acquisition system. Robot control provides active and passive therapeutic exercises.**

I. INTRODUCTION

The use of robots in the ankle-foot neuro-rehabilitation for impaired subjects has to meet challenging requirements in terms of compatibility between the movements allowed by the machine and the physiological movements of the foot. The most prominent factor for such compatibility is the fine alignment between human and robotic articulations, mostly intended as rotational axes, in order not to induce unwanted internal forces. Notwithstanding several available robotic devices, improvements in state-of-the-art solutions are pursued in clinical practice because of some limitations in existing ankle-foot rehabilitation machines. Among them, some can be classified as "wearable devices" [1], [2], [3]. Such devices, equipped with one to three actuators (see Figs. 2-(a),(b)), usually allow a good control of dorsi-flexion, while the tibial internal/external rotation and the foot inversion/eversion are usually kinematically coupled by the machine structure, *i.e.,* combined movements are non-purposefully caused by the device motion. In addition, the position and alignment of actuators can produce internal forces in the ankle, which are not directly related to actual movements and mislead the patient's proprioception. Such drawbacks could severely hinder the rehabilitation process through a deceptive effect on brain plasticity [4]. Finally, wearable devices require an accurate ergonomic design of the necessary orthoses, which should be light, easy to wear and fix onto the wider possible range of anatomical gradations. Another class of devices, "non-wearable" robots, are as much common as the wearable ones [5], [6]. Some of them allow the control of tibial internal/external rotation [7] or adopt the well-known Gough-Stewart platform [8], although it is an over-actuated solution (note that in clinical protocols practiced on a regular basis, translations are typically neglected, due to hip and knee movements: relevant foot motions require only three rotations proper [9]). However, the real main drawback of the majority of"non-wearable" devices is the position of their instantaneous center of rotation right under the foot sole, inevitably distant from the actual ankle articulation (see Figs. 3), and, therefore, unnatural physiological proprioception could be experienced by patients. Notably, the solution proposed in [10] sets a center of rotation located in the ankle center, but they prefer design a two degree of freedom with tibial internal/external rotation coupled with foot inversion/eversion.

Interestingly, some devices, developed for very different original aims and application fields, display many desirable features for ankle-foot rehabilitation. Among others, a simple and promising device is the Agile Eye [11] (see Fig. 4). This fully parallel spherical robot encompasses three parallel legs, each composed of two links, connected to a mobile platform. All the joints are rotational and their axes intersect in the same point, which happens to be the mobile platform rotation center. Although such $(\underline{R}RR)_3$ structure has been designed for industrial applications (motion of lightweight camera systems), few modifications easily allow its usage for ankle-foot motor function rehabilitation.

The paper presents a redesign of Gosselin's robot that fits most of ankle-foot rehabilitation requirements. The prototype, named PKAnkle (Parallel Kinematic machine for Ankle rehabilitation, Fig. 1), is a compact device suitable for sitting/lying patient training and encompassing almost all the physiological range-of-motion of the ankle. The paper describes mechanical features as well as the control functionalities, together with an overview of its design methodology.

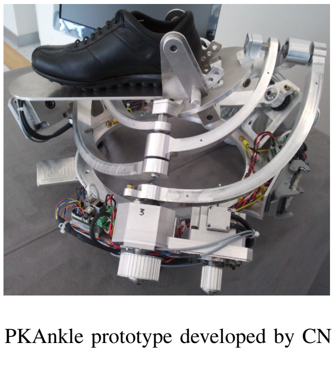

Fig. 1. PKAnkle prototype developed by CNR-ITIA.

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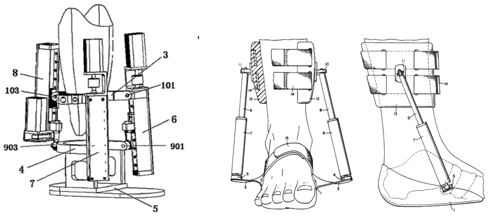
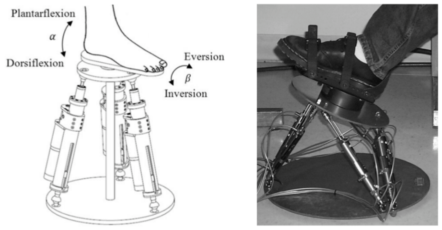


(a) 3-dof parallel device [3]. (b) 2-dof parallel device [1].

Fig. 2. Wearable devices, state-of-the-art



(a) 2-rotational dof more 1 translational dof [1]. (b) 6-dof device, Rutgers Ankle [8].

Fig. 3. External devices, state-of-the-art

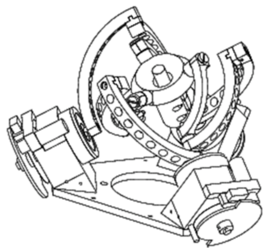


Fig. 4. Gosselin's Agile Eye

II. Ankle-foot kinematics and Application requirements

The ankle is composed of the shank, the talus and the foot [9], and often compared to serial manipulator [12]. The ankle-foot mobility can be therefore easily described by three different rotations: internal/external tibial; tibiotarsal dorsi-plantarflexion; subtalar inversion/eversion (see Tab. I). From anatomic point of view, such rotational axes do not perfectly intersect [9], resulting in residual translational movements of the foot w.r.t. the shank [12]. Such displacements can be however considered negligible for rehabilitation purposes considering the small compensatory movements of the shank. Consequently, the ideal center of rotation of the ankle-foot mobility can be approximated as the midpoint of the shortest segment between the subtalar inversion/eversion and dorsi-plantarflexion axes [9] located in the talus medial part. If the center of rotation of any robotic end-effector is accurately made overlap the center of rotation of the ankle, such robotic device is able to allow a correct use of articulations and muscles, limiting compensatory movements. In particular, the patient can trigger movements of the mechanism by exerting single-component torques around the ankle rotational axes, with no additional effort/strain along other directions. In most of available devices, instead, the robot rotation center results in a position that is (sometime significantly) away from the ankle-foot rotation center (*e.g.,* below the plantar arch). As a result, any muscular action of the patient generates a dorsi/plantar-flexion movement and a force along the tibial axis (*e.g.,* by the biceps femoris contraction when sitting), instead of a due torque around an ankle axis.

III. Design and Therapy Functions

The kinematics of the Agile Eye architecture does not depend on the sizes of its links, but only on the alignment angles among its links. The machine can be scaled according to application needs and its final size does not affect the mobility and dexterity of the mechanism. This property is suited for a twin-step design process: (i) an optimization of the topology on the basis of the mobility requirements of the application (*i.e.,* orientations range) and (ii) a definition of the geometry that matches the machine footprint requirements. The mechanical design happens to be a recursive procedure that aims to balance topological figures, and patient's, therapeutic and therapist's constraints (user-constraints hereafter).

In addition to the mechanical figures computation, the control and the integration of hardware/software have been designed for providing extended therapeutic functionalities. Their design has, in fact, been coordinated with bio-engineers and medical doctors.

A. Dimensioning and Workspace Analysis

As demonstrated in [13], the best conditioned kinematics [14] requires that joint-vectors have to be mutually orthogonal, in order to averagely extend such optimal conditioning in the whole workspace[11]. Likewise, the topological optimum is given by $\mathbf{u}_j \cdot \mathbf{w}_j = \mathbf{w}_j \cdot \mathbf{v}_j = 0$, and $\mathbf{u}_i \cdot \mathbf{u}_k = 0, \forall i \neq k$ (see Tab. II). Although this condition is an optimum in terms of dexterity, the design phase specifically considered some primary user-requirements as (i) the sizes of the mobile platform ($\underline{d}_1$ and $\underline{d}_2$ in Tab. III) and (ii) the

TABLE I
Ankle-foot rotations as in [12]

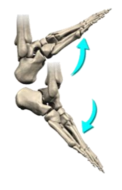
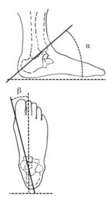
(a) tibial internal/external, (b) tibiotarsal dorsi/plantarflexion, (c) subtalar inversion/eversion	 (a)	 (b)	 (c)
RoM [deg]	[-20, 10]	[-40, 35]	[-25, 20]

TABLE II
Nomenclature

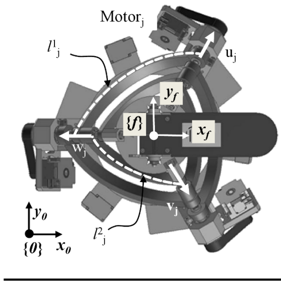
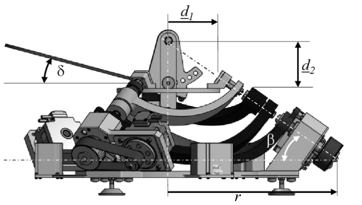
	l^1_j, l^2_j	links of the j-th leg ($j = 1 \dots 3$)
	$\mathbf{u}_j$	axis of the j−th motor
	$\mathbf{w}_j$	connection axis of l^1_j and l^2_j
	$\mathbf{v}_j$	connection axis of l^2_j and the mobile platform
	$\mathbf{q}$	vector of motors coordinates, $[q_1, q_2, q_3]^T$
	$\mathbf{\Phi}$	vector of Rotation of the mobile platform w.r.t base frame {0}
	{0}, {f}	ground and foot (mobile platform) reference frames.

TABLE III

GEOMETRY DIMENSIONING

	User requirements: the dimensions have been fixed on avarage foot dimension, thickness will be used to adapt to different people $\underline{d}_1$ = 110 [mm] $\underline{d}_2$ = 70 [mm]
Design Parameters: β = 35.2 [deg], δ^* = 20 [deg], r^{**} = 238 [mm]	

*minimum angle, mechanical regulation allow to increase it in order to make more ergonomic the movement on the patient.

**depends on links dimensioning.

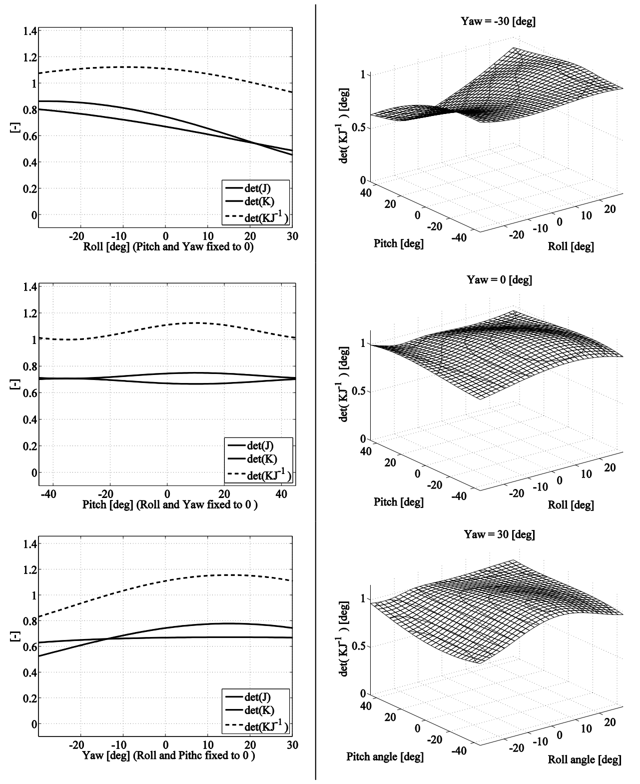


Fig. 5. Denoting $\boldsymbol{\omega}$ as the angular velocity of the mobile platform, the first order kinematic relation is $\mathbf{J}\omega + \mathbf{K}\dot{\mathbf{q}} = \mathbf{0}$. When the determinant of $\mathbf{J}$ or $\mathbf{K}$ degrades to zero (or ∞) singular configurations arise inside the user workspace.

ankle-foot articular range-of-motion (see Tab. I). In addition, the angles $\angle\left(\mathbf{u}_j, \mathbf{w}_j\right)$ and $\angle\left(\mathbf{w}_j, \mathbf{v}_j\right)$ has been constrained to be equal to 90 [deg] because of links machining reasons. Finally, the links l_j^2 are connected to the mobile platform on the points that guarantee the easiest access of the foot and allowing all the links to be under the sole plate when the machine is in its zero-position[1]. Therefore the design procedure involves (i) the angle β of the motor axis with respect to the horizontal plane, (ii) the sizes of the links and (iii) the minimum angle δ of the foot support with respect to the machine plane to avoid collisions inside the workspace (mechanical regulation is allowed to improve ergonomic use by the patient). On the basis of the kinetostatic analysis and in order to guarantee the device manipulability as high as possible, the condition $\mathbf{u}_j \cdot \mathbf{u}_k = 0$, for $j \neq k$ has been adopted during the mechanical design. This condition led to define β angle as reported in Table III. The detailed mechanical design procedure included (i) the definition and numerical assessment of the workspace (kinematics), (ii) the verification of absence of collisions among mechanical parts and (iii) the verification of absence of singular conditions within the workspace boundaries (singular configurations identification has been performed as in [11], see Fig 5). Finally, links and other mechanical parts has been dimensioned according to proper mechanical calculations in realistic working conditions.

Static and dynamic simulations have been performed with two distinct conditions: (i) maximum functional loads in order to derive the maximum motor torques, and (ii) maximum absolute loads in order to derive joints (bearings) and links sizes suited for avoiding any collapse of the structure in case of a patient standing up on the device (see Fig. 6 and Table I).

B. Control Design and Therapy Functions

A key feature in rehabilitation-robotics is the ability of choosing a proper training program/mode based on the specific impairment and recovery status of the patient [6]. The PKAnkle robot controller integrates a force/torque sensor mounted below the sole and an up-to-8 channels EMG signal acquisition card.

1) Controller: it implements three main logics: (i) dynamic compensation in order to allow the machine to be moved directly by patient, limiting the device resistive force; (ii) an admittance-based regulator that partially assists the movement driven by the patient, or alternatively, that opposes a tunable resistance to patient movements; and (iii) a pure position control for passive continuous mobilization (CPM) of the foot. Pure position control can be smoothly switched to an admittance control strategy during the machine movements preserving the control stability. Trajectories can be recorded in dynamic compensation mode, or by scheduling a list of subsequent platform orientations. Motion laws can be independently assigned to each path by associating a velocity profile to the curvilinear abscissa of the preloaded path. In addition, the controller allows the execution of hybrid trajectories constraining the mobile platform of the robot along a given spherical path and letting the motion law to be imposed by the subject (see mode (A) Fig. 7) according to a second order friction model.

2) Therapy functions: controller functionalities can be applied in a wide range of exercises designed by therapists. The separation between path and motion law in the robot interpolator allows setting the velocity as a function of external sensors/needs/wishes. By now, only a velocity inversion algorithm based on the measure of the variation of the ankle impedance during the exercise is implemented in the continuous passive motion.

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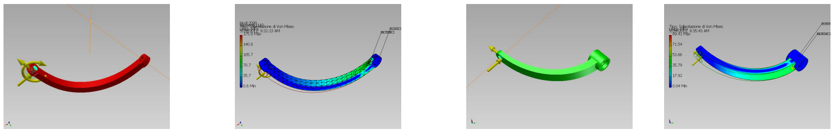
[1] This mounting choice is one of the 8-possible mounting solutions described in [15].

Table IV

FEM LOAD CONDITIONS

	Link 1	*Link 2*
Axial force [N]	243	312
Radial force [N]	312	243
Torque [Nm]	52	0
Material	7075 allumininum alloy (Ergal)	
Young modulus	71	[GPa]
Poisson coeff.	.33	[$-$]
Yield strength.	503	[MPa]



l_j^1 loads l_j^1 Von Mises stresses l_j^2 loads l_j^2 Von Mises stresses

Fig. 6. FEM results: a Force equal to 500 [N] has been applied both on the tip of the foot support and a force equal to 500 [N] has been applied on the heel.

Table V

PKANKLE CONTROL CHARACTERISTIC

Current Loop Rate	3	[kHz]
Position Loop Rate	1	[kHz]
Max. Axis Velocity	90	[deg/s]
Max. Roll Velocity	20	[deg/s]
Max. Pitch Velocity	20	[deg/s]
Max. Yaw Velocity	20	[deg/s]
Continous Axis Torque	6.8	[Nm]
Force Acq. Rate	1	[kHz]
EMG Acq. Rate	1	[kHz]

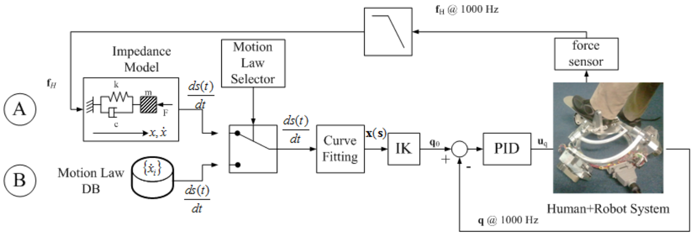


Fig. 7. Robot controller modalities: along a stored path the motion law results from an admittance model (A) or loaded from a database (B). Paths can be recorded directly moving the mobile platform.

IV. HARDWARE DESCRIPTION AND PROTOTYPE EVALUATION

The prototype is built in aluminium, its weight is less than 12 [kg], and the electrical cabinet is easily trasportable. The control software is characterized by an open architecture running on an industrial embedded PC with GNU/Linux and Xenomai real-time patch. Communication among controller, motors drivers and sensors in the machine is provided by an Ethercat master developed by CNR-ITIA. The position control loop rate is 1 [kHz], matching the force/torque acquisition rate. By now, control software implements dynamic compensation and passive continous mobilization. Admittance based control will be delivered in Autumn.

PKAnkle has been preliminarily tested by three healthy subjects. Each subject used the system for about 3 hours, without software/hardware failures, and they have given positive feedbacks on movement smoothness and on the naturalness of velocity profiles. Conversely, the dynamic compensation modality requires additional modeling in order to reduce the motion resistance felt by user.

V. CONCLUSION AND FUTURE WORKS

This paper presents PKAnkle a full-parallel spherical robot for the ankle-foot rehabilitation with a $(\underline{R}RR)_3$ topological structure. The prototype is an adaptation of the Gosselin's spherical robot [11] and the design figures have been determined according to a integrated design process taking into account both kinematic properties and task-requirements. First tests confirm that PKAnkle allows the patient to execute comfortable and physiological movements. Future works will be focused on (i) design a correct chair for an easy access to the device (also for patient in wheelchair), and on (ii) the exploitation of the functionalities allowed by the prototype in order to improve the robot control strategies. Particular focus will be given on the use of EMG signals and Force/Torque measures to modify the robot behavior during exercises.

REFERENCES

[1] J. Bastow, "Ankle rehabilitation device," Patent US 005 215 508A, 06 01, 1993.

[2] J. W. Wheeler, H. I. Krebs, and N. Hogan, "An ankle robot for a modular gait rehabilitation system," in *Proc. of IEEE-SRJ Int. Conf. on Int. Rob. and Sys.*, Sept 28-Oct 2 2004, pp. 1680–1685.

[3] Y. Yin, Y. Fan, C. Guo, Z. Cao, and S. Yin, "Parallel type multi-freedom artificial limb exoskeleton ankle joint," Patent CN 101 234 045, 08 06, 2008.

[4] P. M. Rossini, C. Calautti, F. Pauri, and J.-C. Baron, "Post-stroke plastic reorganisation in the adult brain," *The Lancet Neurology*, vol. 2, no. 8, pp. 493 – 502, 2003.

[5] F. Gao, "Motorized facilitated ankle stretching," in *Mech. and Aut. (ICMA), 2011 Int. Conf. on*, aug. 2011, pp. 102 –106.

[6] Y. Ren, T. Xu, L. Wang, and C. Y. Yang, "Develop a wearable ankle robot for in-bed acute stroke rehabilitation," in *Eng. in Med. and Bio. Soc.,EMBC, 2011*, 30 2011-sept. 3 2011, pp. 7483 –7486.

[7] J. Saglia, N. Tsagarakis, J. Dai, and D. Caldwell, "A high-performance redundantly actuated parallel mechanism for ankle rehabilitation," *Int. J. Rob. Res.*, vol. 28, pp. 1216–1227, September 2009.

[8] J. E. Deutsch, J. Latonio, G. Burdea, and R. Boian, "Rehabilitation of musculoskeletal injuries using the rutgers ankle haptic interface: Three case reports," in *In proc. of EuroHaptics 2001*, 2001, pp. 11–16.

[9] J. Dul and G. Johnson, "A kinematic model of the human ankle," *Journal of Biomedical Engineering*, vol. 7, no. 2, pp. 137–143, 1985.

[10] P. Sui, L. Yao, Z. Lin, H. Yan, and J. Dai, "Analysis and synthesis of ankle motion and rehabilitation robots," in *Rob. and Biomim., 2009 IEEE Int. Conf. on*, dec. 2009, pp. 2533 –2538.

[11] C. Gosselin, E. S. Pierre, and M. Gagne, "On the development of the agile eye," *Rob.& Aut. Mag., IEEE*, vol. 3, no. 4, pp. 29–37, 1996.

[12] C. Syrseloudis and I. Emiris, "A parallel robot for ankle rehabilitation-evaluation and its design specifications," in *BIBE 2008. 8th IEEE Intl. Conf. on*, oct. 2008, pp. 1 –6.

[13] C. Gosselin and E. Lavoie, "On the kinematic design of spherical three-degrees-of-freedom parallel manipulators," *International Journal of Robotics Research*, vol. 12, no. 4, pp. 394–402, 1993.

[14] J. K. Salisbury and J. J. Craig, "Articulated hands: Force control and kinematic issues," *The Int. J. of Rob. Res.*, vol. 1(1), pp. 4–17, 1982.

[15] C. Innocenti and V. Parenti-Castelli, "Echelon form solution of direct kinematics for the general fully-parallel spherical wrist," *Mechanism and Machine Theory*, vol. 28, no. 4, pp. 553 – 561, 1993.