A Soft, Wearable, Quantitative Ankle Diagnostic Device[1]

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1 Background

Approximately 27,000 inversion ankle sprains occur every day in the U.S. [1]. Regardless of the type of sprain, returning to work or sport prematurely is the primary cause for continued ankle disability [1]. Approximately 50% of people who have an acute ankle injury will re-injure the ankle [1]. Therefore, assessment of recovery, including ankle strength and stability, is essential before resuming normal activity. Several studies have shown that leg power and its key components torque and velocity are key indicators of mobility and performance [2–5]. Extending the results from these studies to describe the ankle indicates that measuring ankle power during recovery may lead to a better understanding of the ankle's readiness for work or sport. Although some devices (e.g., dynamometers) allow clinicians to test power and torque capacity, due to factors such as cost and complex protocols, clinicians mostly use qualitative tests such as the drawer, inversion, and star balance test, to determine ankle fitness. As qualitative tests can be difficult to standardize and monitor over time, there is an unmet need to quantitatively assess ankle fitness so that therapists can better understand the condition of the ankle during the recovery period and thus help them decide how best to treat the patient.

2 Methods

Based on discussion with local clinicians and a study of the prior art, we found that measuring ankle power, torque, and angular displacement through the full range of motion could best describe ankle fitness. These measurements must be taken in the frontal plane, since most ankle injuries are related to inversion. As daily life does not typically require maximum torque output, we designed the device to measure up to 30% of maximum ankle torque for an average healthy adult. Other design considerations include a short and intuitive test protocol, ease of donning and doffing, comfort, portability, and cost.

The final design (Fig. 1) consists of a knee brace, sensor bands, a footplate, and user interface. Soft, elastomeric strain sensors (Figs. 2(*b*) and 2(*c*)) are used to exert a torque on the ankle and

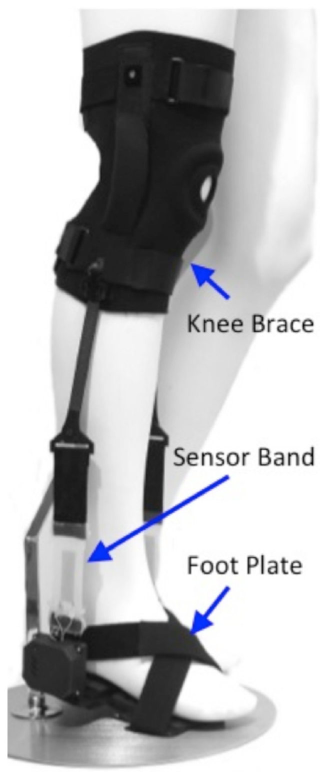

Fig. 1 Final prototype

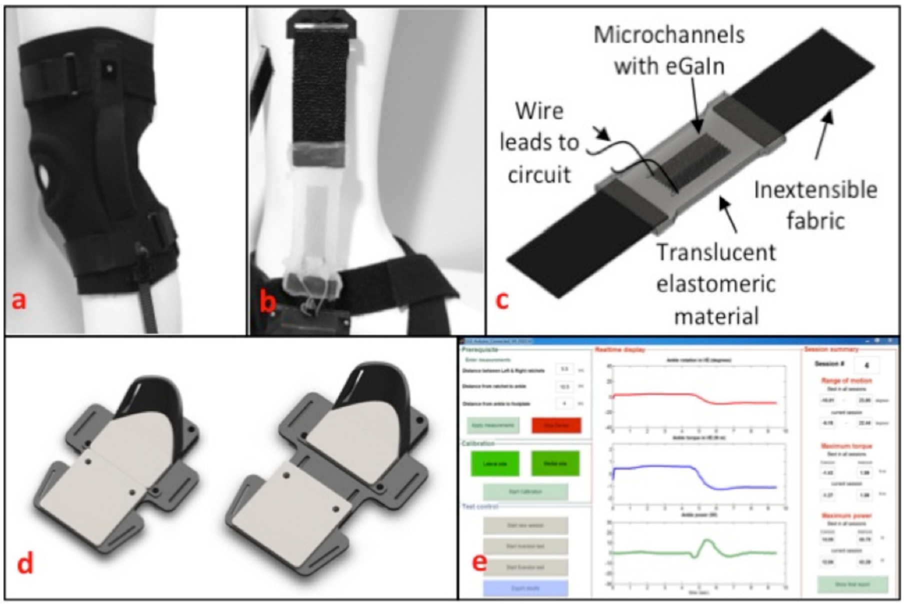

Fig. 2 Device modules: (*a*) knee brace, (*b*) sensor, (*c*) sensor components, (*d*) footplate, and (*e*) user interface

measure angular displacement. When the user inverts or everts, the sensors resist the motion and provide a countertorque about the joint. The sensors were developed by Harvard Microrobotics Laboratory [5] and consist of a translucent elastic material, in this case Dragonskin® 20 by Smooth-On (Macungie, PA), with embedded microchannels filled with eutectic gallium indium (EgaIn). As the sensor deforms, the geometry of microchannels changes, resulting in a change of resistance.

Integrating the sensor into a Wheatstone bridge circuit as a variable resistor gives a voltage output, which corresponds to strain. An Arduino Mega 2560 takes the voltage output from the sensors, and the digitized data from the Arduino are read by a laptop. The laptop postprocesses the data to interpret strain in terms of ankle angle and torque.

The knee brace (Fig. 2(*a*)) provides a consistent attachment point for the sensor bands. We modified a Futuro™ Sport Hinged Knee Brace with velcro straps and linear ratchets to allow for variation in shank length. The plastic strap from each ratchet is attached to a custom buckle, connecting the ratchets to the sensors. Inextensible fabric is embedded on either side of the sensor and sewn to the buckle above and the footplate below.

The footplate (Fig. 2(*d*)) allows ankle motion to translate into a change in sensor length. The sensor bands create a constant moment arm in the inversion/eversion direction through the footplate. As foot length varies from person to person, the plate is adjustable, covering foot sizes from 240 to 290 mm.

The graphical user interface (GUI) (Fig. 2(*e*)) gives clinicians an intuitive way to conduct tests and read results from device. The MATLAB GUI development environment was used to develop the interface, while the SIMULINK Arduino toolkit enables communication between Arduino and the laptop. The analog data are sampled at 100 Hz, and filtered by a second-order Butterworth low pass filter. The cutoff frequency of the low pass filter was selected to 20 Hz, significantly higher than the frequency of movement, around 5 Hz. After filtering the data, ankle angle, torque, and power are calculated with the SIMULINK model.

The overall device can be described by the joint kinematics model (Figs. 3(*a*) and 3(*b*)) and kinetics model (Fig. 3(*c*)), from which one can calculate joint angle, torque, and power based on sensor output. First, the soft sensor (Fig. 2(*b*)) is modeled as a linear spring with a known stiffness value. Therefore, we can measure sensor displacement and force based on voltage output. Second, the sensor displacement is used to measure inversion/

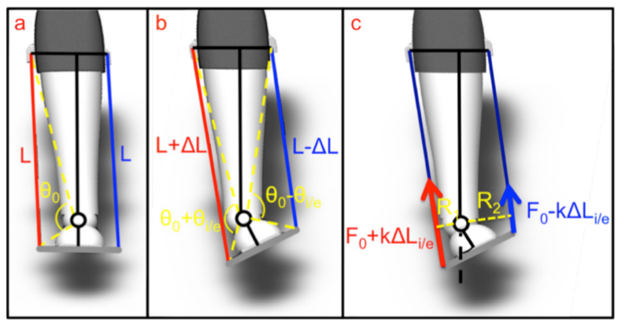

Fig. 3 (*a*) Device model in neutral position, (*b*) kinematics model in inverted position, and (*c*) kinetics model in inverted position

eversion angle based on the kinematics model (Fig. 3(*b*)). We assume symmetry of foot and shank with respect to cross-sectional area parallel to sagittal plane for simplicity. With the kinematic model and premeasured anthropometry such as shank length and width, inversion/eversion angle can be represented by a function of sensor displacements. Third, we can calculate joint torque based on the kinetics model (Fig. 3(*c*)) along with the assumption that inertial force due to foot and footplate mass is negligible. Consequently, the following equation is used to calculate joint ankle:

$$\tau_{\text{ankle}} = R_1\left(F_{\text{pre}} + K\Delta L\right) - R_2 \times \left(F_{\text{pre}} - K\Delta L\right) \quad (1)$$

where R_1 and R_2 are moment arms of sensor with respect to ankle, K is sensor stiffness, and F_{pre} is sensor pretension force. Finally, joint power is calculated based on differentiation of joint angle and torque measurement as shown in the following equation:

$$P = \tau_{\text{ankle}} \times \omega_{i/e} \quad (2)$$

where $\omega_{i/e}$ is ankle angular velocity in *i*/*e* direction. Although the kinetics model is simple and easy to implement in software, the outcome might be inaccurate if the inertial force is not negligible and motion in other plane affects to frontal plane motion.

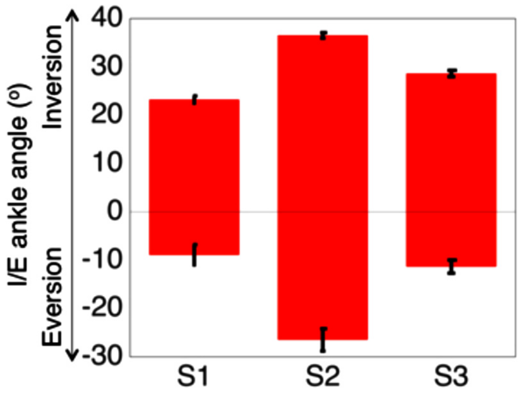

Fig. 4 Range of motion results for three subjects

3 Results

Before being incorporated into the device, the soft strain sensors were calibrated with using an Instron model 5544 A material testing unit in order to determine the relationship between voltage, displacement, and load. The relationship between load and displacement and the relationship between displacement and strain were both approximated as linear curves. However, the prototype circuit used an adjustable gain, which proved to be inconsistent, allowing us to acquire accurate measurements for displacement only. Improving the circuit to a printed circuit board with constant gain should allow us to measure torque and power accurately in the future.

Measurements were taken for three subjects to test device consistency. Four inversion and eversion tests were done for each subject without recalibrating. Results are shown in Fig. 4. The standard deviation of angular displacement in the inversion direction was between 0.58 and 0.76 deg for all of the subjects, while those in the eversion direction were between 1.37 and 2.29 deg. The values are small compared to average range of motion (−35 to +25 deg), which indicates that the device can provide consistent measurements.

4 Interpretation

The quantitative ankle diagnostic device offers a quick and intuitive method for quantitative assessment of ankle fitness while recovering from an injury. While extensive cost analysis has not been done, the cost of the prototype was less than $200, significantly less than current dynamometers and around the same price point as handheld dynamometers, which cannot measure power.

As of now, the device has been tested and evaluated for consistency of measurements for three subjects. Additional trials are necessary to test consistency over multiple testing sessions with separate calibration processes. We also found that we could measure dorsiflexion and plantar-flexion ankle motion, in addition to the original goal of inversion and eversion. A future version of this device may be able to measure angular displacement, torque, and power in both degrees-of-freedom.

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