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Gait analysis for challenged users based on a rollator equipped with force sensors

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***Abstract*— Gait analysis provides insightful information about people condition, progress of rehabilitation treatments and fall risk. This analysis is frequently performed in very specific conditions, because it often implies the use of treadmills, cameras and/or wearable sensors. Alternatively, gait can be analyzed using a smart rollator, equipped with a basic set of sensors. This work proposes a methodology to obtain relevant gait parameters using a simple, cheap source of information: wheel odometry and handlebar force sensors. The main advantages of this approach are that users are not bothered with extra equipment. Besides, monitoring can be performed anywhere, even during everyday conditions and for extended periods of time. We have tested our system with a set of rehabilitation patients with different disabilities. Preliminary results have successfully proven that extracted parameters are coherent with reported effects of their specific condition.**

Fig. 1. Coordinate system in i-Walker platform

I. INTRODUCTION

Assistive robotics improve the quality of life and autonomy of people with disabilities [1]. They also provide a helpful tool for monitoring users with different diseases [2]. Only in United States, 56.7 million people had some kind of disability in 2010 [3]. Of these people, 11.76 million have some mobility challenge.

Smart rollators are a useful support for Activities of Daily Living (ADL) in users with mobility challenges [4] and they have a significant role in rehabilitation [2], [5]. These rollators have a double purpose: support and monitoring. Rollators can support up to 64% of the vertical load [6]. Also, it is an effective walking aid for partial weight bearing, which is essential in the healing period of orthopaedic lower limb patients [5]. Additionally, they provide a useful tool for fall prevention and gait monitoring [7].

Gait monitoring is crucial in patients with neuropathology (stroke, hemiplegia, ataxia, Parkinson, etc) to measure gait parameters as walking velocity or stride length [8]. Furthermore, it is essential in orthopaedic lower limb patients to monitor partial weight bearing [5] and also in older adults to predict potential falls [9].

Gait parameters are typically measured on treadmills [10], using cameras [11], electronic walkway [12], or wearable sensors [13]. Unfortunately, gait in rollator users may be conditioned by treadmills [14]. Besides, monitoring with cameras or electronic walkways need to be performed in a controlled area (e.g indoors environments [11]). Finally, wearable sensors are often not comfortable enough for long term use [13]. Also, since they need to be adapted for different users to be worn, testing with several users may take significantly longer.

Smart rollators do not condition users' gait as long as they are used passively. Besides, they can be used in everyday conditions and for long term monitoring. These rollators have integrated sensors that may spare the user from the burden of wearables. Also, they are suitable for a variety of users with minimal configuration changes (handlebar adjustment to height, if necessary). Although there are parameters that can only be measured with external sensors, like hip rotation or tibial torsion, smart rollators offer a good alternative for monitoring during ADL. Nevertheless, they can measure postures as hip rotation or tibial torsion using a external sensors only (cameras, wearable sensors, etc).

[15] presents a cheap and simple solution for monitoring based on a 3-wheels rollator equipped simply with odometry and a gyrometer. Sway motion during walking is typically related to a force pair in the rollator handlebars that results, specially in a 3-wheels rollator, in a minor rotation on the frame. This swing can be extracted from gyrometer and analyzed along with odometry to extract a number of parameters. Unfortunately, tests in [15] were limited to users with no pathological walking diseases and the obtained result in terms of step length differs significantly from other studies [16], [12].

In 4-wheeled rollators, gyrometer changes related to sway are not so easy to capture. Besides, gyrometer is compromised when the user is not following a straight line. In

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this work, we propose to use force sensors in the rollator handlebars to directly obtain the force pair instead of estimating it from odometry. This approach provides more accurate measures and additional parameters, plus it is valid for any type of rollator.

The purpose of this work is, consequently, to develop a methodology to monitor gait parameters using a smart rollator equipped with force sensors in the handlebars. The main contributions of this work are: i) definition of meaningful parameters related to medical conditions and rehabilitation progress; ii) a simple and cheap procedure to extract those parameters from a smart rollator; and iii) validation with a set of users presenting a variety of cognitive and/or physical disabilities. The main advantage of this proposal with respect to previous work is that: i) it is valid for any environment; ii) it requires no wearables nor cameras; and iii) it has been validated by the target users. Although this is a preliminary study with a limited number of volunteers, results have been successful. We expect to use the proposed methodology to evaluate the users rehabilitation progress and to warn about potential falls.

II. Gait

The ability to walk upright is a defining characteristic of humans. Cognitive and/or physical disabilities may produce gait abnormalities. A classification of gait abnormalities in the elderly can be found in [8]. These abnormalities can often be characterized by quantitative gait parameters. Monitoring of parameters like step time or weight-bearing can be useful for diagnosis support, evaluation of progress or risk prediction. For example, users with antalgic gait are expected to improve weight-bearing through rehabilitation [17].

There are several studies on relevant parameters for different disabilities. We have compiled the most popular ones in Table I. It shows how gait abnormalities affect gait parameters in healthy users: increasing (↑), decreasing (↓), not affecting them (−) or increasing their variability (∨). As suggested in [12], the average user has been chosen in the age interval 70-74. This average is set by healthy users walking without aid. Clinicians in our team reported that healthy individuals do not use walkers correctly, so it would be unrealistic to work with them.

If an assistive device can capture a set of these parameters, it could provide an estimation about the user's condition/evolution in an unsupervised fashion and even adapt assistance accordingly. For realistic estimation, it would be desirable to capture these parameters in everyday situations, keeping the user's load and wearable devices to a minimum.

A. Estimation of gait parameters using a sensorized smart rollator

A smart rollator equipped with force sensors in the handlebars provides continuous information on how the user distributes weight. Fig. 2 shows forces on both sides for a user with prosthetic left femur fracture by time. We appreciate a fluctuation in the right force and in the left force,

TABLE II

Acronyms, units and definition

Acronyms	Units	Definition
CAD	$\frac{steps}{min}$	Cadence
SpT	s	Step Time
SpL	m	Step Lenght
SdT	m	Stride Time
SdL	m	Stride Lenght (Stance phase)
WV	$\frac{m}{s}$	Walking Velocity
WB	N^{-1}	Weight-bearing

almost in antiphase[1]. Phase shifting is die to regular motion sway: when a person initiates heel contact, the handlebar force in the same side increases and the handlebar force in the opposite side decreases [30]. Hence, we can detect the user's steps by simply searching for inflection points in the function resulting from difference between forces $f_{diff} = F_Z^{right} - F_Z^{left}$ (fig. 3(g)).

Besides, we can extract several gait parameters in Table I from f_{diff}:

- Step time (SpT): Average time between maximum-minimum (Right) or minimum-maximum (Left) in seconds.
- Stride time (SdT): Average time between maximum-maximum (Right) and minimum-minimum (Left) in seconds.
- Number of Step (NoS): Numbers of inflection points.
- Time required (Tr): Number of seconds that the user takes to complete the test.
- Cadence (CAD): $60 * \frac{NoS}{Tr}$.

As commented, most smart rollators provide odometry. We can obtain further gait parameters in Table I using odometry calculation. Function f_d represents walked distance in time (fig. 2). With this function we measure:

- Step length (SpL): Average length between maximum-minimum (Right) or minimum-maximum (Left) in seconds.
- Stride length (SdL): Average length between maximum-maximum (Right) and minimum-minimum (Left) in seconds.
- Distance(d): Distance walked by user in meters.
- Average walking velocity (WV): $\frac{d}{Tr}$.

User's supports (UrS) can be estimated as the sum of forces $F_Z^{left} + F_Z^{right}$ for all left or right steps. When a rollator supports much weight, the user's lower extremity in stance phase is supporting less weight. Hence, we can measure weight-bearing (WB) as the inverse of UrS

III. Methodology

A. Our platform: the i-Walker

The i-Walker platform [31] is a smart robotic rollator based on an standard MEYRA® walker frame. Each handlebar sensor has 3 force components. It includes encoders in both wheels. Besides, it also includes a 2D laser, a tilt sensor and 2 forces sensor to measure the normal in each hind legs.

[1]The amplitude difference is due to the specific condition of the user: she tends to avoid putting too much weight on the left leg

TABLE I
HOW DISABILITY AFFECT TO GAIT PARAMETERS

Gait	Division	Gait Abnormalities						
		CAD	SdT	SdL	SpT	SpL	WV	WB
Healthy [12]	Men Women	102(8) 113(20)	1.18(0.08) 1.06(0.13)	1.39(0.014) 1.23(0.17)	0.59(0.05) 0.53(0.06)	0.69(0.08) 0.61(0.09)	1.17(0.16) 1.16(0.2)	– –
Antalgic	affected non affected				↓ [18] - [18]	↓ [18] - [18]		↓ [19]
Ataxic	-	↓ [20]	∨ [21]	↓ [21]	↑ ∨ [21]	↓ [21]	↓ [21]	
Hypokinetic		- [22]		↓ [23]		↓ ∨ [21]	↓ [21]	
Vestibular		↑ [24]	↓ [24]				↓ [25]	
Spastic		↓ [26]					↓ [26]	
Paretic	affected Non affected	↓ [27]	↑ [27]	↓ [27]	↓ [28] - [28]	↓ [27]	↓ [27]	↓ [28]
Cautious [29]	-			↓		↓	↓	
Dyskinetic	Involuntary movements or postures, these abnormalities can not be measured consistently.							

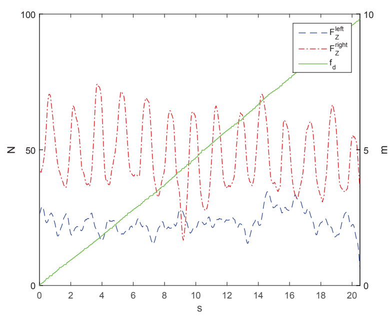

Fig. 2. Handlebar forces (F_Z^{right}, F_Z^{left}) by time (User 7)

Indeed, we use in combination with odometry only 1 force component in the handlebar sensors to obtain all commented gait parameters: the Z component (fig. 1). This component measures the user's vertical load support.

Force sensors have a sampling rate of 20 Hz and a resolution of 0.98 N. This provides high resolution for the analysis of user data. Errors due to resolution limitation may affect the WB parameter. They do not affect spatiotemporal parameters because f_{diff} during walking is always larger than 2x0.98 N for people who need rollators to move.

SpT and SdT are affected by the accuracy to determine when user initiate heel contact. The relationship between gait parameters and forces in handlebars on rollators has an accuracy $67.26 \pm 50.38ms$ when users move in a straight line [30]. In addition, SpL, SdL, d and WV parameters include an odometry error $3.53 \pm 0.0068\%$ on one meter[2].

B. Users: Patients from Hospital Regional Universitario of Malaga

Testing devices with people with disabilities is usually challenging, so many works in this field tend to work with healthy volunteers ([11], [6], [30]). Other works focus on a very specific kind of user for monitoring gait parameters (e.g. parkinson [2], stroke [10] or ataxia [20]). Our target is to define a more general methodology for users with disabilities who require the use of a rollator. Hence, we have worked with volunteers from the rehabilitation unit of Hospital Regional Universitario de Malaga. We have tried to work with a variety of conditions. We imposed only three restrictions: volunteers had to: i) present a mild to severe disability profile; ii) be able to walk with the aid of a rollator [32]; and iii) have experience with rollators, to avoid cold start related issues.

In this work we present data from 9 volunteers: 6 women and 3 men. They had a variety of cognitive and/or physical disabilities (table III). Users are in average 68.22 ± 14.63 years old (range $45 - 86$ years).

C. Test: the 10-meter walk tests

There are many tests to measure gait [33]. However, the relationship between gait parameters and forces in handlebars can be measured with better accuracy when users move in a straight line [30], since the force pair in handlebars is not affected by rotation maneuvers. Rotation effects can be reduced from force signals if we use odometry records for correction. However, in order to prove that gait parameters can be measured via handlebar forces and are actually related to the user's condition, it is better to stick to simple tests involving walking a straight line to keep signals are clean as possible. Therefore, we have a limited number of options. For our first approach, we used the inexpensive 10 mWT test [33]. It consists of walking 10-meters in a straight line. This test presents an number of advantages. It is a widely proven test, valid for different diseases and with reported successful results [34], [33]. Furthermore, it is a frequent test during the different stages of clinical rehabilitation. Hence, it can be easily integrated into clinical sessions at hospitals and it is easier to obtain approval from their Ethical Committee.

The 10 mWT test has two part. First, the user walks 10 meters in a straight line[3] at a comfortable pace. Afterwards, the user walks the same distance at his/her maximum safe pace. In this work, the second part of the test was not approved at the Hospital because we had no previous results with the i-Walker yet and it involves higher fall risks. Therefore, all results refer to the first part of the test.

IV. RESULTS

The goal of this section is: i) to show that we can extract all commented gait parameters using only the i-Walker in

[2]We measured the odometry error by walking in straight line 1 meters 20 times to simulate a stride length.

[3]Users actually walked 12 meters, but we discarded the first and last meter to eliminate acceleration and deceleration effects.

TABLE III

USERS CONDITION

User	Age	Gender	Physical	Neurological
1	69	Women	Right Side afected	Stroke
2	45	Women	Polytraumatism in both lower limbs	Psychological distress
3	54	Women	Tetraparesis	Low grade astrocytoma
4	59	Men	Spinal fusions. Left leg rotated	-
5	86	Men	Intertrochanteric hip fracture (Left)	Psychological distress
6	59	Men	Above knee amputation (Left)	-
7	77	Women	Prosthetic femur fracture (Left)	-
8	83	Women	Total hip replacement (Left)	-
9	82	Women	Intertrochanteric fracture femur (Left)	-

TABLE IV

GAIT PARAMETERS BY USERS

User	Leg	CAD	SdT	SdL	SpT	SpL	WV	UrS
1	Left Right	73.9336	1.5813(0.4308)	0.7649(0.2002)	0.8458(0.2896) 0.7308(0.3655)	0.4302(0.1412) 0.3296(0.171)	0.4829	56.5301(7.0931) 55.4994(7.8742)
2	Left Right	33.6347	3.619(0.5296)	0.6473(0.1369)	1.8733(0.3698) 1.74(0.3781)	0.3183(0.1328) 0.329(0.0522)	0.1801	141.7732(50.2771) 138.5881(58.2236)
3	Left Right	55.5556	2.187(0.1967)	0.7089(0.0618)	1.125(0.0541) 1.0708(0.184)	0.3339(0.0541) 0.3806(0.0546)	0.3248	107.3875(32.7621) 112.432(28.6415)
4	Left Right	51.1013	2.3741(0.2923)	0.6933(0.0459)	1.0821(0.2366) 1.3036(0.2341)	0.3208(0.0465) 0.3731(0.0482)	0.2899	184.1019(39.8449) 193.2380(32.2076)
5	Left Right	142.5532	0.8408(0.0956)	0.2039(0.0253)	0.4076(0.0911) 0.4348(0.0701)	0.1029(0.0222) 0.102(0.0221)	0.2423	129.4412(10.126) 124.7827(10.1273)
6	Left Right	34.1969	3.4935(0.6364)	0.6256(0.1848)	1.6813(0.4557) 1.8156(0.4456)	0.376(0.1817) 0.2465(0.043)	0.1782	181.2906(56.7767) 122.3216(44.4457)
7	Left Right	81.95	1.5058(0.1728)	0.72(0.0791)	0.6357(0.1486) 0.8654(0.1819)	0.3150(0.0721) 0.4018(0.0786)	0.4788	73.0059(11.5508) 70.9994(12.8941)
8	Left Right	82.3713	1.4262(0.192)	0.2242(0.0468)	0.7143(0.1898) 0.7091(0.1931)	0.1124(0.0331) 0.1103(0.0356)	0.1544	217.7974(50.0625) 228.4787(70.0204)
9	Left Right	72.7273	1.6192(0.2324)	0.5846(0.0931)	0.7932(0.2231) 0.8214(0.1578)	0.2845(0.0853) 0.3005(0.0676)	0.3626	59.062(7.8287) 57.6729(8.3495)

passive mode; ii) to show that parameters behave as reported in table I; and iii) to check that parameter values are coherent with each volunteer's diagnosed condition.

Table IV shows the gait parameters measure per user. We worked with challenged users, therefore their walking velocity were very slow compared to healthy users (table I). As expected, results matched each user's pathology. Next, we extend our comments to specific cases.

A. Ataxic gait

User 1 presented ataxic gate, typically provoked by a stroke [19]. Her right side was physically affected by the stroke. She was at the end of her rehabilitation process when we did our tests. We can see in Fig. 3(a) how f_{diff} is noisier around right heel peaks. This happens because she had difficulty to load weight on her right side. This causes, as expected for this gait, abnormalities, a higher variability in SdT (0.4308) and SpT (0.3655) and a lower value of CAD (73.9336), WV (0.4829), SdL (0.7949) and SpL (0.3296). Her weight-bearing was high (UrS below 60 N) because at the end of the rehabilitation process, she had already regained much of her strength .

B. Antalgic gait

Antalgic gait is provoked by pain or a limited range of motion [8]. It can be observed in users 2 (both limbs affected) and 4 to 9 (hip or femur affected or amputation). This gait abnormality is characterized by a limp, provoked by the affected side who has a lower value for step time and length. We can check, for example, user 7, with a prosthetic left femur fracture. Her SpT values were 0.6357 and 0.8654, for left and right side, respectively, since she purposefully avoided leaning on her left side. This effect is also clearly appreciated in her higher left UrS value (fig. 3(g)). However, some users often avoid pain and the related limp effect by increasing their weight support in the rollator. User 8 was an example of this behavior (fig. 3(h)). She had a total hip replacement, left side affected. We can observe in table IV how her SpT values for left and right side were quite similar (0.7143 vs 0.7091). However, her UrS was larger than 200 in both cases, well over the rest of our volunteers and almost 4 times larger than users like 1 or 9. Due to her enormous UrS, this volunteer managed to walk just 5 meters before she grew too tired to continue. User 9 and 5 were at the end of their rehabilitation process, so the difference between sides for SpT (less than 0.03 s) and SpL (less than 0.02 m) were reduced. User 2 had both limbs affected, therefore she does not have much difference between SpT (1.8733 vs 1.74) and SpL (0.3183 vs 0.329). But, we can observe the effect of pain in the higher variability in these parameters SpT (0.3698) and SpL (0.1383) (fig. 3(b)).

C. Vestibular gait

User 5 also presented vestibular disease, according to his obtained parameters. This user was very unstable because he could not maintain a stable posture with respect to the rollator. Indeed, he tended to load more weight to right. We can appreciate in figure 3(e) how he started exerting more force on the left at the beginning, but he clearly favored his right side towards the end. As in vestibular gait, he kept a high CAD (142.55) and a low SdT (0.8408) (fig. 3(e)).

D. Cautious gait

Cautious gait is related with fear to fall [8]. User 6 walked very slowly (fig. 3(f)), almost 2 times slower than user 1, 3, 7 or 9. Also, he had lower SdL (0.6256 vs 1.39) and SpL (0.376 vs 0.69) values. These gait parameters correspond to a cautious gait. We observe also how his UrS was almost 70% bigger in left side, this was because he was learning how to walk with prosthetic leg and he had fear to fall.

E. Paretic gait

Paretic gait can be associated with weakness on one side (hemiparesis) or both side (tetraparesis). SdT for user 3, was almost 2 times larger than in healthy users (2.187 vs 1.06) and she had lower values of CAD (55.5556 vs 113), SdL (0.7089 vs 1.23) and SpL (0.38 vs 0.61) (fig. 3(c)). These parameters coincide with her tetraparesis, where her right side was more affected.

V. Conclusion and Future Work

In this paper we have presented a new methodology to measure gait parameters in people with disabilities using a rollator equipped with force sensors and odometry. Unlike usual methodologies, that require treadmills, controlled areas or specific wearable sensors, our system works does not condition gait because it relies on a device familiar to users: a rollator. Indeed, our rollator can be used to carry ADL and for extended periods of time.

The main limitations of our method are the following ones. Unlike more complex methods, it does not provide enough information for detailed gait analysis (e.g. kinematic analysis of joint rotations). Besides, it requires the user to lean on both handlebars while walking.

Our methodology has been validated with a number of volunteers presenting a variety of disabilities. Tests have successfully proven that: i) the proposed methodology returns the most frequent parameters to evaluate gait analysis; ii) those parameters evolve according to reported clinical studies; and iii) resulting gait, as characterized by our results, is coherent with the reported users' diagnosis. Hence, these results validate our proposal to analyze gait with a smart rollator equipped with force sensors and odometry.

Future work will focus on normalizing the obtained data using the approach presented in [35]. These data from users at the end of the rehabilitation process will be stored in groups presenting similar gait abnormalities to obtain a stereotype user. This stereotype will be used to compare the rehabilitation process of a new user with the same gait abnormalities. We also plan to apply our methodology to a set of users during their entire rehabilitation process to obtain a objective measure of their progress with respect to their rehabilitation treatment. We expect that results from such work will allow us to adapt assistance to each person using the i-Walker on a need basis by acting on the device motors (shared control).

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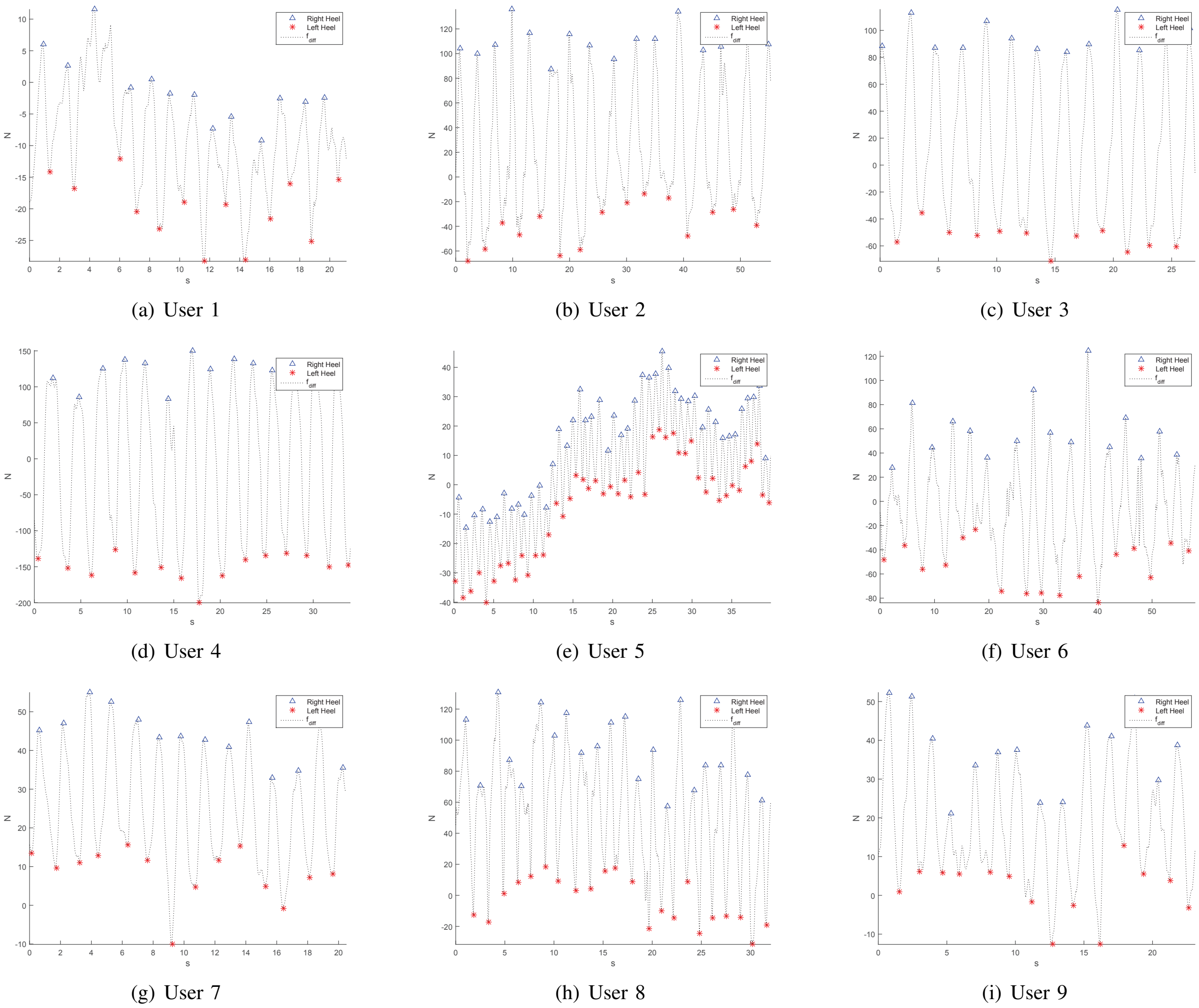

(a) User 1 (b) User 2 (c) User 3

(d) User 4 (e) User 5 (f) User 6

(g) User 7 (h) User 8 (i) User 9

Fig. 3. f_{diff} in some users

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