mobileSpiro: Accurate Mobile Spirometry for Self-Management of Asthma

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Abstract

Effective management of asthma and other respiratory diseases requires constant monitoring and frequent data collection using a spirometer and longitudinal analysis. However, even after three decades of clinical use, there are very few personalized spirometers available on the market, especially those connecting to smartphones. To address this problem, we have developed mobileSpiro, a portable, low-cost spirometer intended for patient self-monitoring. The mobileSpiro API, and the accompanying Android application, interfaces with the spirometer hardware to capture, process and analyze the data. Our key contributions are automated algorithms on the smartphone which play a technician's role in detecting erroneous patient maneuvers, ensuring data quality, and coaching patients with easy-to-understand feedback, all packaged as an Android app. We demonstrate that mobileSpiro is as accurate as a commercial ISO13485 device, with an inter-device deviation in flow reading of less than 8%, and detects more than 95% of erroneous cough maneuvers in a public CDC dataset.

Categories and Subject Descriptors

J.3 [**Computer Applications**]: Life and Medical Sciences—*Cellular computing*

1 Introduction

Asthma is a major problem globally, especially among children [1]. One of the biggest challenges for asthmatics is preventing a serious attack in which the patient lung capacity is significantly diminished, leading to severe shortness of breath and emergency treatment. In most cases, the degradation of lung capacity occurs over a prior period of two to four days and is detectable *if* patient's lung capacity can be measured daily.

Such daily measurements can be taken using a spirometer, which also provides many indices of respiratory health. However, even with more than three decades of clinical use and numerous studies demonstrating that daily spirometry readings can predict impending asthmatic episodes accurately [2], there are almost no spirometers in the market for patients to monitor themselves.

One reason for lack of patient-operated spirometers is the high chance of error in performing a test. The forced expiratory maneuver, a standard test in spirometry, appears quite straightforward. In this maneuver, the patients inhale to completely fill their lungs and then exhale as forcefully as possible for as long as possible into the spirometer. However any error in the maneuver, such as less-than-maximal effort, invalidates the reading. Additionally, erroneous readings can lead to false alarms and unneeded visits to the doctor.

In this paper, we present mobileSpiro for performing spirometry at near-lab accuracy by end users. There are three main components of mobileSpiro: (a) custom spirometer hardware which is small-factor, low-power and very low-cost to enable an affordable end-product, (b) a custom API to allows any Android device to interface with the spirometer via Bluetooth, allowing for development of patient-care applications and (c) robust error-detection algorithms enabling effective patient self-monitoring.

Our hardware solution is as accurate as a commercially available spirometer (all priced at more than $500); our custom hardware component costs are no more $100 in large quantities. In addition, our error-detection algorithms can detect over 95% of coughs and 74% of early termination maneuvers. Such automated error detection allows for higher quality in-lab spirometry measurements and enables patient self-monitoring.

The rest of the paper is organized as follows. We will first describe the current state of asthma monitoring in Section 2. Next, we detail the mechanics and possible errors of a forced expiratory maneuver (Section 3). Lastly, we illustrate how our architecture addresses the spirometry maneuver errors (Section 4) and demonstrate its effectiveness in Section 5.

2 State of the Art

Our work is inspired by the paradigm that health management is a shared responsibility of the patients and the care provider. Remote monitoring technology in general has been shown to improve patient outcomes, because the patients themselves gain greater knowledge about what symptoms they need to take care of.

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In [3], the authors demonstrated that electronic journaling, where the asthmatic patients have to choose one of several options, leads to improved compliance by patients in tracking their health condition. In addition, it was shown in [4] that spirometry self-testing through an Internet environment is comparable to validity as a technician-monitored spirometry test session. The experiment was conducted on subjects with no computer background, showing that this technology is usable by people of all backgrounds.

However, these encouraging results have not translated into the larger practice of self-management of asthma at home. Currently, children with moderate to severe asthma rely on infrequent visits to a physician, typically in the range of two to four times a year [5]. However, the physician does not know of the child's status in between visits unless something major happens to the child, such as a severe asthma attack resulting in an emergency room visit. One of the major reasons is lack of cost-effective and personalized asthma management tools [6].

With mobileSpiro, we plan to address all the major shortcomings in developing an end-to-end solution. mobileSpiro's combination of low-cost, accurate hardware, software with automated patient feedback, a journaling system and open-source API ensures high-quality spirometry data and enables new applications for user engagement.

3 Spirometry Basics

Spirometry is the most common way to measure lung capacity of a patient. Breathing at rest does not utilize the full capacity of the lungs; at-rest tidal volume is limited to a small fraction of total capacity. As the lungs are not at their limit, declining lung function symptoms are not noticed until there has already been severe degradation. Spirometry is meant to highlight the decline in function by forcing the patient to expel the maximal volume possible with maximal effort, allowing doctors to detect the early onset of both restrictive and obstructive diseases.

3.1 Spirometry Maneuver

In the forced expiratory maneuver, the patient inhales as much as possible and then exhales into the spirometer as hard as possible for as long as possible. Throughout the maneuver, the output flow of air and the total volume of air exhaled thus far is measured and recorded. In addition to providing many predictive indices of respiratory health, such data can be used to construct a flow vs. time curve useful for diagnosing respiratory diseases.

The correct maneuver (Figure 1) is characterized by:

- Beginning the maneuver with maximal blast effort
- Applying maximum effort throughout the maneuver
- Keeping a tight seal on the spirometer mouthpiece to avoid leaks
- Avoiding variability in output flow, such as breaks
- Completing the maneuver in one continuous breath

The acceptability of a maneuver is determined not only by the shape of the spirogram but also by certain parameters obtained from each maneuver (Figure 2). These include:

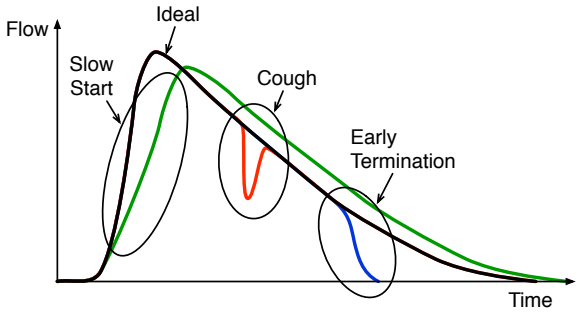

Figure 1. Ideal maneuver with possible errors

- FVC (Forced Vital Capacity): the total volume of air exhaled during the expiratory phase of the maneuver.
- FEV1 (Forced Expiratory Volume After One Second): the total volume of air exhaled one second into the maneuver.
- PEF (Peak Expiratory Flow): the maximum recorded flow during the course of the maneuver.
- FEV1/FVC Ratio: the ratio of the FEV1 to the FVC of any maneuver. An acceptable ratio is above 0.85.

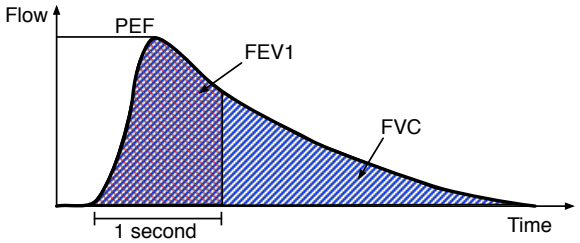

Figure 2. Spirogram parameters annotated on a sample graph

A correct maneuver requires maximal effort, a supportive environment, and most importantly a good state of mind. Ideally, the patient assumes the same position in all maneuvers; standing is preferred. The patient must wear loose clothing so that their lungs and airways are unconstricted. Lastly, patients should be not allowed to smoke or eat within an hour of performing a maneuver.

3.2 Erroneous Maneuvers

Although spirometry provides powerful diagnostic indices, the slightest mistake can invalidate a spirometry maneuver. The following errors are the most common and important errors identified in spirometry tests.

Cough: When a patient coughs into the flow tube, the aberration caused is very prominent in the spirogram. The detection of a cough in the spirometry maneuver is based on a simple principle: there should be one peak flow which is both maximum locally and globally. That is, the shape of the spirometry curve should contain only one sharp peak. The flow should be strictly increasing until the single peak expiratory flow is reached; afterwards, the flow should be strictly decreasing. Therefore, the detection of cough is based on the extent to which these conditions are defied. A comparison of

a cough maneuver with the ideal is shown in Figure 1.

Hesitation and Slow Start: In this error, the patient does not start with the maximum blast effort. In the spirogram, the rise of volume with respect to time is not steep enough. To check this, if the peak expiratory flow (PEF) occurs at a volume larger than 0.7 L, hesitation occurred [7]. Slow starts are also bound to occur when there is excessive extrapolated volume. These two conditions go hand in hand, and the presence of either indicates a slow start. An example maneuver is shown in Figure 1.

Early Termination of Maneuver: A maneuver is said to be incomplete if one of two conditions are met: glottis closure or short-lived maneuver. Glottis closure, as the name implies, occurs when there is complete loss of flow when the glottis or epiglottis suddenly closes. This appears as a plateau in the volume or a sudden drop in flow to zero. Short-lived maneuvers are maneuvers that do not last the full required six seconds; in this case, the patient ceases expiratory effort before the six second mark. Both glottis closure and short-lived maneuvers are seen in Figure 1.

4 mobileSpiro Architecture

The mobileSpiro architecture has three main components. The first is the hardware, which samples, collects, and filters the raw data for transmission to the Android device. The second is actual recording of the spirometry maneuver, in which raw sampled data is converted into flow measurements as per the calibration. Lastly, the third is the high-level spirometry maneuver analysis system, where the recorded spirometry maneuver is processed by validation algorithms to determine which errors, if any, are present in the maneuver. The overall system is depicted in Figure 3.

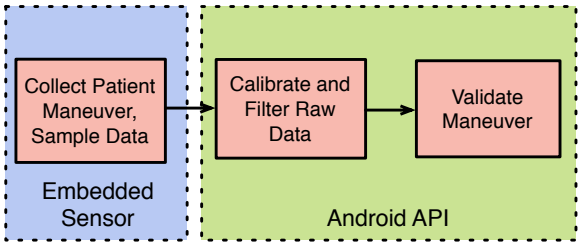

Figure 3. System Architecture

4.1 Hardware Design

Our approach was to leverage existing technologies to encourage adoption. Smartphones are rapidly pervading all levels of society and have the necessary computing power to process raw data and run data analysis algorithms. Therefore, respiratory monitoring would simply require a low-cost sensor. The hardware architecture is shown in Figure 4.

The spirometry hardware is responsible for data collection. To maximize patient convenience, the two design goals were (1) low-power consumption and (2) portability. We chose to use replaceable batteries as the power source, so that the end-user would not have to wait for any recharging. The actual hardware is shown in Figure 5; the black box is the custom spirometer, and the white tube is the laminar flow tube, which reduces the turbulence of the air flow.

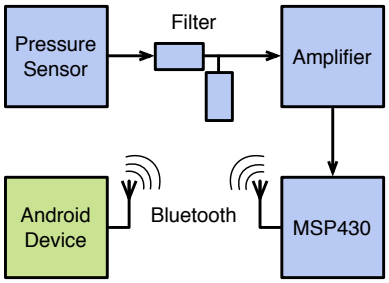

Figure 4. mobileSpiro Hardware Architecture

At the front end for data capture is a pneumotachometer, which converts a stagnant pressure input to an analog output. Our hardware uses a Freescale MPXV7002 which has a range of ± 0.7 psi. The sensor is bidirectional, enabling the capture of both exhalation and inhalation flows. The analog output varies between 0 V and 5 V with zero pressure at 2.5 V.

The maximum input bandwidth is no more than 1 kHz; to filter out noise components, a simple RC-filter is implemented after the input. The components (750 Ω, 0.33 μF) have a cutoff frequency of 1 kHz. The pressure sensor range is ± 0.7 psi, which is larger than needed for a human spirometry maneuver; therefore, an amplification factor of 6 is applied prior to digital sampling.

A Texas Instruments MSP430F5437 samples the data at 1 kHz and transmits the raw samples to the Rayson BTM-182 Bluetooth module. The Bluetooth module is a UART connection at 115 kBaud. The 12-bit sampled data is sent as two separate words with the first 4 bits of the first word indicating a data byte. The data received at the smartphone is processed by low-level Linux drivers as part of the Android Bluetooth stack. The software checks for the 4-bit sequence in alternate receptions to ensure synchronization with the custom hardware. The data is made available to the calibration and validation subsystems at the higher layers; Section 4.4 describes the Android processing stack in detail.

Figure 5. Actual mobileSpiro Hardware: The mobileSpiro processing block and laminar flow tube.

All parts used in the spirometer are standard off-the-shelf components, with higher tolerances, significantly reducing the cost of the design. Also, the PCB design is compact, further reducing the assembly cost.

The input to the pressure sensor is an off-the-shelf laminar flow tube. The turbulent airstream is laminarized by the

tube to attain a linear relationship between pressure and flow. It creates a stagnant pressure column based on the flow rate; thus, the pneumotachometer output is directly proportional to the input flow rate. Hardware calibration is performed to ensure accuracy of flow rates using the methodology mentioned in [8]. The procedure involves linear regression with data from a 3 L calibrated syringe. Over 20 plunges of the syringe at slow, medium and fast flow rates are collected and used to determine a third-order mapping of digital samples to calibrated flow rates.

Power Utilization: The custom hardware is powered by two AA batteries. The batteries output a voltage of 3.1 V when charged and 1.8 V when drained. There is a boost regulator which generates 5 V and 3.3 V for the digital system. As per the datasheets, the Bluetooth module is the dominant power sink in the system, drawing up to 60 mA when transmitting. The pressure sensor draws up to 10 mA, and the MSP430 requires less than 1 mA when running in low-power mode.

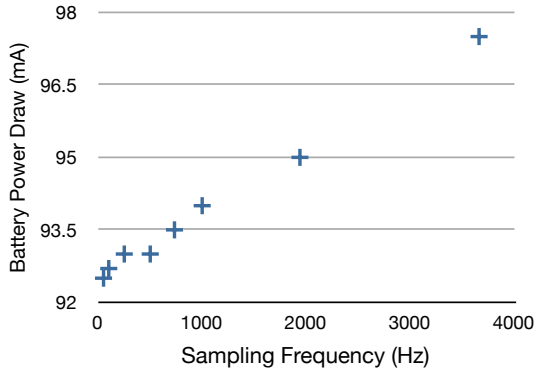

Figure 6. Power Consumption of Spirometer Hardware

The actual power utilization of the system is shown in Figure 6. The graph plots power draw from the AA batteries with varying sampling rates of the data. As sampling rate increases, the power consumption increases, as the MSP430 must remain awake longer. However, this incremental decrease is small compared to the total power draw. As the Bluetooth module must remain active during data collection, it dominates the power consumption.

For a sampling rate of 1 kHz (the upper limit expected by ATS), the draw is 94 mA. If the device is used twice daily for up to 10 minutes each, standard 2500 mAh batteries will last for up to 80 days.

4.2 Software Design

The spirometer software is a pure Java API (shown in Figure 7), built for use on the Android platform. It includes software packages useful for interacting with the spirometer hardware and for analysis of spirometry maneuvers.

The first layer of the API stack is the collection of data from the sensor. Because Android devices have differing availability of Bluetooth, mini-USB, and microphone input, the data service is interchangeable. In addition to the hardware RC filter, each implementation of the top-level spirometer interface defines its internal own digital filter as well. Essentially, the digital filter is responsible for translating the

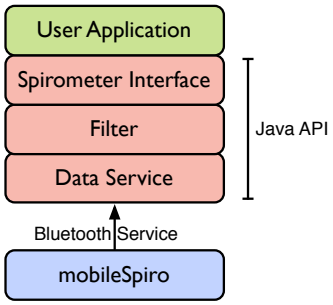

Figure 7. Java API

incoming raw pressure values, which represent only relative quantities, into calibrated, meaningful flow values.

The top layers of the API stack focus on storage and analysis of the obtained spirometry maneuver itself. Each spirometry maneuver can be stored and associated with patients in a built-in SQLite database. On the analysis end, important modules include the interchangeable validation module, which defines the strictness of validation of maneuvers as well as the algorithms used to evaluate the maneuver. Such flexibility is necessary for adapting the spirometer for use among different demographics; for example, it is recommended that children be held to less rigorous standards in validating acceptable spirometry maneuvers [9]. Lastly, applications can register themselves as listeners on the spirometer input, allowing for seamless integration with the hardware. The end result is a cohesive package that enables rapid development of Android applications making use of the mobileSpiro hardware. We profile our own development of an asthma journaling application, which makes full use of the mobileSpiro API.

4.3 Spirometry Validation Algorithms

In a clinic, technicians or doctors can monitor spirometry maneuvers to ensure their accuracy. In contrast, when patients operate their personal spirometer, we open the door to erroneous maneuvers which do not accurately reflect the patient's lung function. To address this problem, we implemented real-time algorithms in the software, which validate each maneuver. For each maneuver, we check for the following errors: a cough, hesitation, glottis closure, and short-livedness. Each of these errors manifests itself in the flow-time and volume-time curves.

Cough: As seen from the graphs in Section 3.2, a cough is characterized by a sudden drop succeeded by a rise in flow. The cough detection algorithm works by finding all local maxima in the maneuver. Because there is only one peak (the peak expiratory flow) in the flow-time curve of an ideal spirometry maneuver, the presence of more than one local maximum indicates that a cough occurred. If a local maximum other than the PEF exists and differs from the PEF by more than 0.6 L/s, then the algorithm detects a cough. The 0.6 L/s value was determined experimentally based on observed noise levels in our hardware.

Early Termination of Maneuver: In a short-lived maneuver, the flow drops to zero before 6 seconds after the peak expiratory flow is reached. However, as discussed in [9], that

Error	Feedback
Cough	"Try to avoid coughing."
Hesitation	"Start faster, without hesitation: exhale as forcefully as possible from the start."
Short-lived maneuver	"Keep on blowing air out."
Glottis closure	"Try again; keep on blowing air throughout the entire maneuver."

Table 1. Feedback from Validation System to User

is often difficult for children, so we use 5 seconds. This is checked by looking for the last group of meaningful values above 0.5 L/s (to avoid errors due to noise). If that group of values is less than five seconds from the start of the maneuver, it is classified as short-lived. If the slope of values dropping off is greater than 70 degrees, it is classified as a glottis closure.

The performance of these validation algorithms is studied in Section 5.2.

4.4 Patient-feedback System

The software is intended to allow the patient to accurately self-track symptoms and monitor their asthma, in conjunction with a physician monitoring their performance. When the new test subject is entered into the Android software, the patient is greeted by two options: whether to take a new test or to view their recent history. Currently, the patient is stored locally on a database, but server-side implementations would allow for greater flexibility in patient data access. The data capture and validation algorithms are both packaged into an Android app distributed with the hardware.

When the patient chooses to take a new test, an automated spirometry test session is launched. When the software has established a connection with the sensor, the message "begin when ready" is displayed, prompting the patient to begin the spirometry maneuver at his or her discretion. The software displays a real-time graph of flow and volume versus time, which serves to motivate the patient to give his or her best effort. Real-time displays of peak expiratory flow and forced vital capacity are also displayed as the maneuver progresses. Up to a third of a second's worth of data before the beginning of the maneuver is also logged in order to preserve the maneuver's entirety, depending on how quickly the patient begins the maneuver.

Lastly, when the software has detected the end of the maneuver (defined here as 1 second of flows between ± 0.1 L/s, due to the presence of noise), the software logs the data and then brings the patient to a new screen where they can review the results of their test. On the left, the patient feedback section is displayed. Should an error be detected, an appropriate coaching message is displayed, and the patient is given the option of re-doing the test. For example, if a cough was detected, the software displays "Try to avoid coughing". Such coaching messages are described throughout spirometry literature, and we have chosen the ones most suitable for patient self-monitoring [7]. All the feedback messages are shown in Table 1.

Figure 8. Screenshot of Feedback View

During the "review test" stage, checkboxes display a set of common asthma symptoms, allowing patients to quickly and accurately describe their respiratory condition. Capturing this metadata is extremely valuable, as declines in lung function involve both a qualitative increase in symptoms and a change in quantitative factors, such as a decrease in peak expiratory flow over time. The patient can also write notes about their test.

On the recent history view screen, several status indicators immediately indicate the current performance of the patient. One of the most important indicators is the rolling 15-day history graph. Depending on the patient's recent performance in spirometry tests, the graph is shaded with a different intensity of either green, yellow, or red. The performance of the patient is based primarily on the current model of peak flow monitoring, that of being in the green, yellow, or red zone [2]. The software informs the patient which zone they are in and can also prompt the patient to take steps based on their action plan. In the future, the performance of the patient could be categorized using more sophisticated measures or integrated with other medical sensors to provide a more personalized portrait of current health.

Since poor performance in any test may result from any number of factors, ranging from simple operator error to symptoms of asthma, the patient is given the option either re-doing the test or saving the test result if any error is detected in the maneuver. However, if the test is saved and contains any error, the maneuver will be flagged appropriately if submitted for later, external review.

5 Full System Evaluation

The hardware and validation algorithms are evaluated for their accuracy against well-known systems. First we test the output accuracy of the hardware against an ISO13485 approved device. To test the validation algorithms, we run the same algorithms against the CDC released NHANES III [10] spirometry data set. The database has been annotated by both a technician and a computer so we can benchmark our validation performance against those results.

5.1 Hardware: Sensor Accuracy

To determine the accuracy of the sensor, we benchmarked our calibrated mobileSpiro hardware with the ISO13485-certified Thor PC FlowMeter. By comparing the flow values reported from the FlowMeter and our spirometer over a range of flows, we were able to determine the

	P_Det	P_FP
Cough	95.5%	15%
Early Termination	74%	14%

Table 2. Algorithm Evaluation

accuracy of our mobileSpiro sensor values as compared to an established commercial spirometer.

In this experiment, the air source was a pressure pump; the air was piped from the pump to the laminar flow tube through a long coil of tube intended to laminarize the flow. Because the air from the pressure pump itself was quite turbulent, this coil of tube was necessary to control the flow's degree of turbulence. By adjusting the pressure of the pump, it was possible to vary the speed of air flow through the spirometer. For each different flow supplied by the air pump, we alternately fitted our hardware and the Thor PC FlowMeter to the end of the long coil of tube. Therefore, the flow value was kept constant, and we were able to compare the accuracy of our calibration versus a commercial medical device.

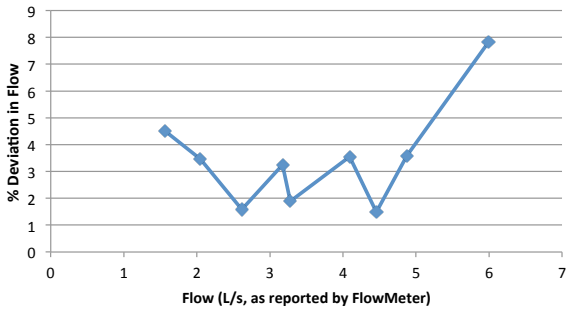

Figure 9. Results of benchmarking experiment: deviation from FDA-approved device versus the flow rate.

In Figure 9, we show the maximum inter-device deviation in reported flow values. With increasing flow rate, the deviation between our custom hardware and the FlowMeter never exceeds 8% across 0 L/s to 6 L/s. Because the control to the air pump was very coarse and the variability of flow increased as the flow increased, we were unable to obtain higher resolution in our benchmarking.

5.2 Algorithms: Accuracy of Error Detection

In this experiment, spirometry datasets were taken from the NHANES III database [10], a rich source of spirometry information. The raw curves were extracted from the database file and run through the validation algorithms. Each NHANES curve is annotated with a technician and computer score on the validity of the maneuver. Table 2 summarizes the results.

Cough detection: For cough detection, all maneuvers which were tagged as a cough by the technician and by the NHANES computer software were extracted from the NHANES III dataset. These maneuvers were run through our validation algorithm. 95.5% of the maneuvers were accurately detected as coughs. The algorithms also had a 15% false positive rate for the NHANES maneuvers that were marked as good.

Early Termination: Selecting the subset of the database from the NHANES with the 'early-termination of expiration' error reasoning from the NHANES computer, our early termination algorithms successfully detected 74% of the errors, while the false positive rate is just 14%.

We tried to validate the algorithms for hesitation as well but due to inconsistent labeling of the NHANES III dataset, we were unable to extract the maneuvers that correspond to hesitation in reality.

Overall, the software algorithms substitute well for a technician in the self-monitoring scenario.

6 Conclusion

In this paper we presented mobileSpiro, a new portable smartphone-connected hardware spirometer. To alleviate the requirement of a trained technician, the mobileSpiro automated system checks the maneuver quality as the data is collected. Our algorithms successfully identify over 95% of coughs as tagged by a CDC database. Additionally, we can successfully capture early-termination maneuvers. All this ensures that longitudinal data collected daily by the patient is accurate and can be used for clinical diagnosis.

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