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Design & Implementation of Covid-19 Emergency Ventilator

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ABSTRACT: Amid the global crisis caused by the corona virus pandemic, hospitals and healthcare facilities are reporting shortages of vital equipments. One important device here is ventilators for patients who need assistance with their breathing due to the respiratory effects of COVID-19.A DIY ventilator may not be efficient as that of a medical grade ventilator but it can act as a good substitute if it has control over the following key parameters like Tidal volume, BPM, IE Ratio, Flow rate, Peep. This design is based on the automation of the manual BVM (Ambu-bag), a hand-held device used to provide positive pressure ventilation. When the bag is squeezed, air enters the lungs, while the nonreversible breathing valve prevents backfiring of the exhaled air. Then the AMBU bag self-dispenses by sucking air from the valve from its back side. Either ambient air can be used as "fuel", or an oxygen cylinder can be connected. In the latter case, it is possible to connect a tank to collect excess oxygen, which was not used by the patient. Most of the DIY ventilators are based on ambu bags & a direct drive actuator.The key element in my design is a linear actuator which is coupled with a lever mechanism that can compresses the ambu bag. A control panel is also provided for precisely controlling the ventilation parameters.

KEYWORDS: Ventilator, Ambu bag, Tidal volume, Flow rate, BPM (Breathe Per Minute), IE Ratio, PEEP, PIP.

I. INTRODUCTION

Coronavirus disease 2019 is an infectious disease caused by severe acute respiratory symdrome coronavirus 2. It was first identified in December 2019 in Wuhan, China, and has resulted in an ongoing pandemic. The first case may be traced back to 17 November 2019. As of 8 June 2020, more than 6.98 million cases have been reported across 188 countries and territories, resulting in more than 401,000 deaths. More than 3.13 million people have recorded.

There are many mechanical ventilators on the market with different levels of complexity and sophistication. The most sophisticated hospital ventilators have integrated sensors, electronics, and software intelligence that control the volume of air flow, air pressure, and breathing rate. Because of the exponential growth of COVID-19 infected patients, it is currently unknown whether enough hospital-grade ventilators will be available to meet demand in the coming weeks and months. When a hospital-grade ventilator is not available,alternative ventilation methods are desired.

A low function ventilator is an alternative to a hospital grade ventilator that is simpler, less expensive, and has lower capability for controlling air flow during ventilation. There are a variety of low function ventilators that are commercially available, with different ventilators designed for different application and intended to satisfy the requirements of those applications. An automatic resuscitator is a device that can replace hand-bagging to provide oxygen to a patient who has stopped breathing, a portable ventilator or transport ventilator is capable of ventilation

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outside of a hospital or while the patient is being moved, and a disaster ventilator or emergency ventilator is a device that is deployed in an emergency when there are patients that need a ventilator but a hospital grade ventilator is not available. While low function ventilators all have lower capability than a conventional hospital grade ventilator, they can be attractive for certain situations where the attributes of low cost,simplicity, and accessibility are important. We refer to the device of this study as an emergency ventilator or EV, because of its intended use in an emergency and because we demonstrate ventilation of an animal using the device.

II. SENSING SYSTEM

To ensure that it is broadly useful in as many settings as possible, the sensor and alarm system is not designed to be integrated with any particular model of ventilator. Rather, it is a standalone component that can be attached to any pressure cycled ventilator. Because it is intended to address an emergency shortage, the design prioritizes cost and ease of production. The system uses low-cost, widely available parts, can be assembled on a two-layer printed circuit board using either through-hole or surface-mount components, and runs on a standard 5 volt power supply.

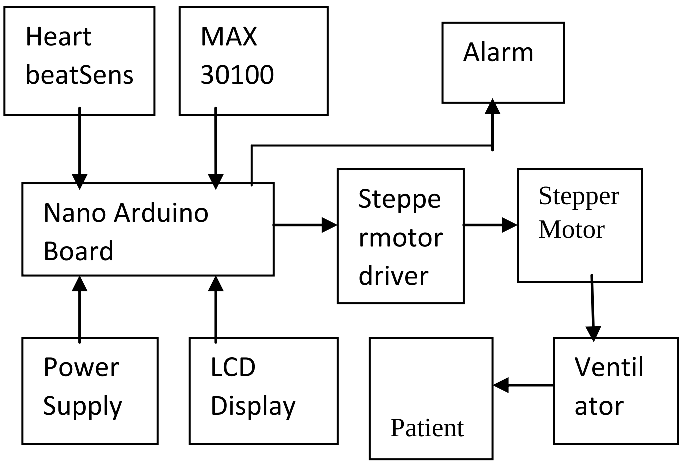

FIG 1: Block diagram of Covid-19 Emergeny Ventilator

The device connects to the patient airway using standard respiratory tubing adapters attached on the patient side of the ventilator. The electronic system consists of a microcontroller, a LCD display, push buttons, a buzzer, and a pressure sensor. Our implementation uses the 8-bit microcontroller ATmega328, which was selected for its ease of use and wide availability. It is driven by an internal 8 MHz clock and does not require an external oscillator.

One important device that is ventilators for patients who need assistance with their breathing due to the respiratory effects of COVID-19. Basically a ventilator is a machine that provides breathable air into and out of the lungs, to deliver breaths to a patient who is physically unable to breathe, or breathing insufficiently. A DIY ventilator may not be efficient as that of a medical grade ventilator but it can act as a good substitute if it has control over the following key parameters

Tidal volume: It's the volume of air delivered to the lungs with each breath by the ventilator - typically 500ml at rest.

BPM(Breaths per minute): This is the set rate for delivering breaths.

Inspiratory:Expiratory ratio (IE Ratio): refers to the ratio of inspiratory time:expiratory time.

Flow rate: is the maximum flow at which a set tidal volume breath is delivered by the ventilator

Peep (Positive end expiratory pressure): It is the pressure in the lungs above atmospheric pressure that exists at the end of expiration.

My design is based on the automation of the **manual BVM (Ambu-bag),** it is a hand-held device commonly used to provide positive pressure ventilation.

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When the bag is squeezed, the air enters the lungs of the patient, while the nonreversible breathing valve prevents backfiring of the exhaled air. Then the AMBU bag self-dispenses by sucking air from the valve from its back side. Either ambient air can be used as "fuel", or an oxygen cylinder can be connected. In the latter case, it is possible to connect a tank to collect excess oxygen, which was not used by the patient.

Most of the DIY ventilator out there are based on ambu bags and a direct drive actuator. In my design, I've tried to simplify the actuator mechanism and I also made a better user interface for the unit.

The key element in my design is a linear actuator which is coupled with a linear mechanism that can compress the ambu bag. A control panel is also provided for precisely controlling the ventilation parameters.

In order to make a linear actuator, I used a bipolar stepper motor integrated with a lead screw(or a Nema 17+coupling) and coupled it with a traveling nut. As the shaft of the motor rotates the traveling nut translates along the screw so that linear motion is achieved.

In the end, control over ventilation parameters can be achieved by varying;

No. of rotations (stroke) ----> Tidal volume

Speed of rotation----->Flow rate

Steps in clockwise : steps in anti-clockwise direction--->IE Ratio

Frequency of direction change per minute---->BPM

I'm using an **A4988** for controlling the stepper motor. The A4988 is a microstepping driver for controlling bipolarstepper motors which has built-in translator for easy operation. This means that we can control the stepper motor with just 2 pins from Arduino, ie one for controlling the direction of rotation and the other for controlling the steps.

For user interface I've made a control panel out of a 20X4 character display and 3 buttons (for up, down & ok functions). You could simply connect the display directly with arduino but I'd prefer using an **I2c display adapter** so that you can plug and play without a mess. **10K** resistors are added for each individual buttons for pull down purpose.

III. PRESSURE-CYCLED VENTILATION

Pressure-cycled pneumatic ventilators, which are powered by pressurized gas, are useful in the present emergency because they have low cost, are easy to manufacture, and require no electronic components for basic operation. They provide pressurized gas to the patient airway and cycle between inhalation and exhalation modes using a pressure-switching mechanism controlled by pneumatic logic, as shown in Fig. 2. During inhalation, high-pressure gas flows from the ventilator to the patient's lungs. As the lungs inflate, the pressure in the airway increases until it reaches the peak inspiratory pressure (PIP), a maximum pressure threshold that can be adjusted by the user. Once PIP is reached, the modulator opens a path to the atmosphere that allows air from the lungs to exit the ventilator. During exhalation, the pressure in the airway drops steadily, but does not fall to atmospheric pressure. Instead, once it drops below the positive end-expiratory pressure (PEEP), a spring closes the path to the atmosphere to initiate the next inhalation. During assisted-breathing mode, also known as pressure-support mode, the patient initiates a breath by inhaling to pull the pressure below the PEEP threshold. The clinician can control flow by adjusting a PIP dial and a rate dial, which determines expiratory time.

In pneumatic ventilators, the PEEP threshold is a fixed fraction of PIP determined by the mechanical design of the device. Because COVID-19 patients can require PEEP levels in the range 10–15 cm H_2O and PIP levels in the range 30–40 cm H_2O [4], some COVID-19 emergency ventilators are designed with smaller PIP-to-PEEP ratios than commercial ventilators. For example, the Illinois RapidVent has a measured PIP-toPEEP ratio of about 2.4

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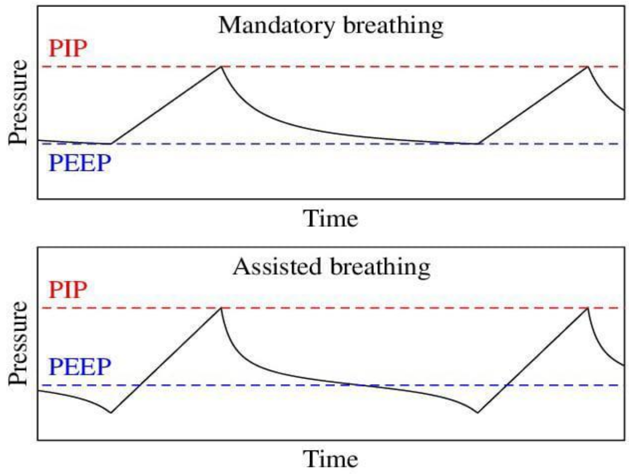

Fig 2: A pressure-cycled ventilator uses positive pressure to deliver gas to the patient airway. During normal operation, it produces a distinctive pressure waveform.

IV. PRESSURE TRACKING

The behavior of pressure-cycled ventilators is well characterized by the pressure signal measured at the patient airway. During normal operation, the pressure cycles between PIP and PEEP once per breath, as shown in Fig. 2. In an ideal system, breaths could be tracked by simply finding maxima and minima in this signal. However, real signals do not always increase and decrease monotonically like the waveform in that figure. The tracking algorithm must be robust against small pressure variations and must have low computational requirements so that it can run on inexpensive, low-power microcontrollers.

Alarms

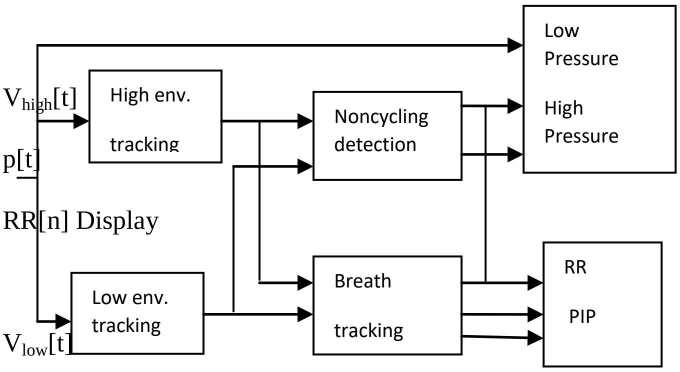

Fig 3: Alarm conditions and clinical metrics are derived from the measured pressure signal.

The proposed processing system, illustrated in Fig. 3, uses a pair of nonlinear recursive filters to track the envelope of the pressure signal. Recursive filters of the form $y[t] = ay[t-1] + bx[t]$ are widely used in signal processing for their computational efficiency: because they use feedback from the output to the input of the filter, they can perform many filtering tasks with less memory and fewer multiplications than feed-forward filters.

Let $p[t]$ be the discrete-time pressure signal from the sensor, where t is the sample index. The high-pressure envelope $v_{high}[t]$ and the low-pressure envelope $v_{low}[t]$ are given by

$$v_{high}[t] = \begin{cases} \alpha A V_{high}[t-1] + (1-\alpha A)p[t], & \text{if } p[t] \geq V_{high}[t-1] \\ \alpha R V_{high}[t-1] + (1-\alpha R)p[t], & \text{if } p[t] < V_{high}[t-1] \end{cases} \quad (1)$$

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$$v_{low}[t]= \begin{cases} \alpha A V_{low}[t-1] + (1-\alpha A)p[t], & \text{if } p[t] \le V_{low}[t-1] \\ \alpha R V_{low}[t-1] + (1-\alpha R)p[t], & \text{if } p[t] > V_{low}[t-1] \end{cases} \quad (2)$$

where $\alpha A \in [0, 1]$ and $\alpha R \in [0, 1]$ are called the attack coefficient and release coefficient, respectively. These coefficients determine the relative importance of old and new samples in the moving average; they control how quickly the envelope tracker responds to changes in the pressure signal.

V. VENTILATION MONITORING

The monitoring system estimates three metrics: PIP, PEEP (or the minimum pressure of the breath cycle for assisted breathing), and RR. All three of these metrics are tracked by detecting inhalation and exhalation cycles from the pressure envelopes. The two envelope trackers each store the most recent pressure sample that triggered their attack mode, as shown in the figure. During each inhalation cycle, there are several attack-mode samples in a row for the high-pressure envelope. During exhalation, there are several attack-mode samples in a row for the low-pressure envelope. The system tracks breath cycles by looking for low-pressure attack events that follow high-pressure attack events and vice versa. A low-pressure attack event causes the system to switch from inhalation to exhalation mode, and a high-pressure attack event causes it to switch from exhalation to inhalation mode.

PIP and PEEP

When a mode switch occurs, the previous attack value is used to update the corresponding PIP or PEEP estimate. That is, when a low-pressure attack event occurs, the PIP display is updated with the most recent high-pressure attack value. When a high-pressure attack event occurs, the PEEP display is updated with the most recent low-pressure attack value. Let $V_{high}[n]$ and $V_{low}[n]$ be the peak values of the high- and low-pressure envelopes, respectively, during breath cycle n, and let $T_{high}[n]$ and $T_{low}[n]$ be the sample indices at which they occur.

Both PIP and PEEP are recursively smoothed over time to remove small fluctuations:

$$PIP[n] = \alpha S PIP[n - 1] + (1 - \alpha S)V_{high}[n] \quad (3)$$

$$PEEP[n] = \alpha S PEEP[n - 1] + (1 - \alpha S)V_{low}[n], \quad (4)$$

where αS is a smoothing coefficient between 0 and 1. This is a linear filter with an exponential impulse response; the contribution of sample n0 to the moving average decays as $\alpha(n-n0)$ S . The closer αS is to 0, the more quickly the display will respond to changes. We used $\alpha S = 0.5$ in our implementation.

Respiratory Rate

The system also keeps track of the time elapsed between these mode-switch events. A complete breath cycle is measured between high-pressure peaks. After smoothing, the average number of samples per breath is

$$Tb[n] = \alpha S Tb[n-1] + (1 - \alpha S)(T_{high}[n] - T_{high}[n-1]) \quad (5)$$

Then the respiratory rate in breaths per minute is given by

$$RR[n] = 60fs /Tb[n] \quad (6)$$

where fs is the pressure sensor sample rate in samples per second. Note that although they are described mathematically as signals, in practice T_{high} and T_{low} are implemented as counters that reset on each breath cycle.

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TABLE I
ALARM CONDITIONS

Alarm	Condition	Tunable range on RapidAlarm
High pressure	$p[t] > p_{\max}$	$30 \leq p_{\max} \leq 90$ cm H_2O
Low pressure	$p[t] < p_{\min}$	$1 \leq p_{\min} \leq 20$ cm H_2O
High RR	$\mathrm{RR}[n] > \mathrm{RR}_{\max}$	$15 \leq \mathrm{RR}_{\max} \leq 60$ breath/min
Low RR	$\mathrm{RR}[n] < \mathrm{RR}_{\min}$	$5 \leq \mathrm{RR}_{\min} \leq 15$ breath/min
Noncycling	$t - T_{\text{high}}[n] > T_{\max}$ $t - T_{\text{low}}[n] > T_{\max}$ $v_{\text{high}}[t]/v_{\text{low}}[t] < r_{\min}$ $v_{\text{high}}[t] - v_{\text{low}}[t] < d_{\min}$	$5 \leq \frac{T_{\max}}{f_s} \leq 30$ sec

VI. ALARM CONDITIONS

The monitoring device triggers alarms in several conditions that indicate the ventilator is not working properly, as shown in Table I. The alarm thresholds may vary between patients and between ventilator devices and so they are configurable by the user.

Pressure and Respiratory Rate

The high- and low-pressure alarms trigger immediately if the sensor detects a pressure outside the permitted range. In a pressure-cycled ventilator, the pressure should never exceed the PIP value set by the user. A pressure reading above the range of the PIP dial indicates a mechanical failure. The low-pressure threshold pmin can be set close to zero, that is, atmospheric pressure, to detect a disconnect in the breathing circuit. Note that because pressure-cycled ventilators apply positive pressure even during exhalation, the airway should never drop to atmospheric pressure unless the patient is attempting to breathe spontaneously.

The high and low-respiratory-rate alarms trigger if the average respiratory rate falls outside the range specified by the user. A high respiratory rate could indicate a low tidal volume, for example due to deteriorating lung compliance, that requires a clinician's attention. The low-respiratory-rate alarm has some overlap with the noncycling alarm, but it triggers based on the average time between complete breath cycles, while the noncycling alarm is triggered by the time elapsed since the last breath event.

Noncycling Conditions

The noncycling alarm condition is more complex than the first four. It must detect when the breathing cycle has stopped, which can happen in several ways. Thus, the alarm can be triggered by several conditions.

First, the alarm triggers if too much time has passed since the last attack event of either envelope. For example, if the pressure drops to PEEP and remains constant, as shown in the top panel there will be no attack events in the high-pressure envelope tracker, so it will trigger the alarm. If, however, the pressure fluctuates slightly over time, as shown in the bottom panel, the tracking algorithm will still detect frequent peaks.

To handle this case, the alarm will also trigger if the highpressure envelope and low-pressure envelope are too close together. In pressure-cycled ventilators, the ratio between PIP and PEEP is a constant, here denoted rnom, determined by the mechanical design of the device. For the Illinois RapidVent, the nominal ratio is around 2.4. An alarm is triggered if $v_{high}[t]/v_{low}[t]$ drops below r_{min}, a pressure-ratio threshold between 1 and r_{nom}. The alarm also triggers if the difference $v_{high}[t] - v_{low}[t]$ is too small. In our implementation, this minimum difference is fixed at 3 cm H2O.

VII. RESULT

Arduino Nano setup connected to the ventilator to check the abnormal condition of the patient.

In pneumatic ventilators, the PEEP threshold is a fixed fraction of PIP determined by the mechanical design of the device. Because COVID-19 patients can require PEEP levels in the range 10–15 cm H2O and PIP levels in the range 30–40 cm H2O [4], some COVID-19 emergency ventilators are designed with smaller PIP-to-PEEP ratios than commercial ventilators. For example, the Illinois RapidVent has a measured PIP-toPEEP ratio of about 2.4

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Parameters	Normal	Abnormal
Flow rate	High	Low
Tidal Volume	500	1500
BPM	72	15
IE Ratio	1:2	6:6

VIII. CONCLUSION

In response to the COVID-19 pandemic, we designed, prototyped, and tested an emergency ventilator. The design of the ventilator is based on a continuous flow, Pressure Limited Time Controlled format and the parameters of the device are tested in accordance with the Emergency Use Authorization guidance. The proposed sensor and alarm system can improve the functionality of pressure-cycled emergency ventilators.. The goal of this device is to allow a patient to be treated by a single ventilator platform, capable of supporting the various treatment paradigms during a potential COVID-19 related hospitalization. This test is used to detect risk of lung injury and after a post mortem, the lungs were found to be well protected by the ventilator during this procedure. The ventilator is shown to be a promising candidate for emergency use during the COVID-19 pandemic and beyond in cases where a rapid-response and versatile ventilation platform are needed.

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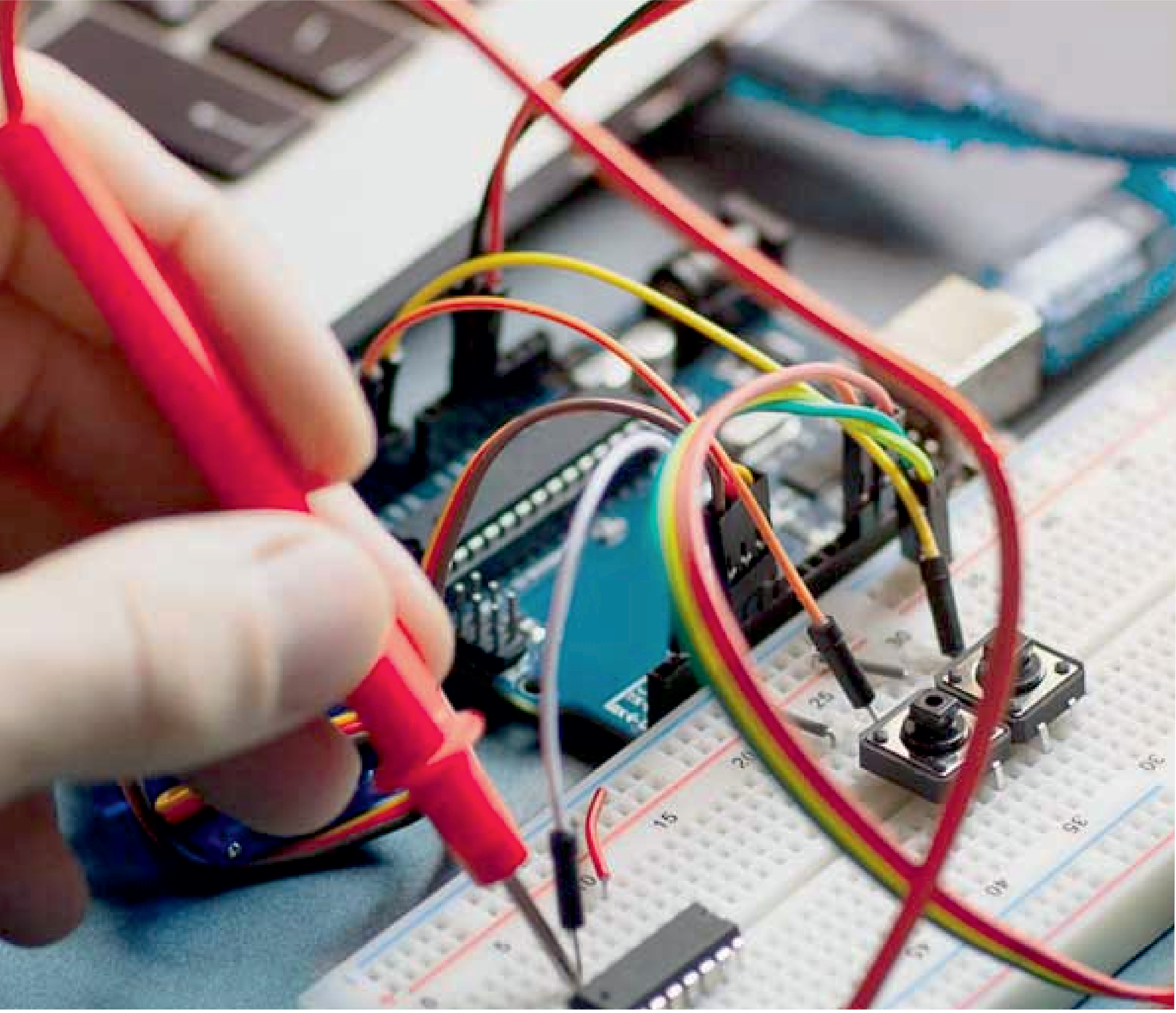

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